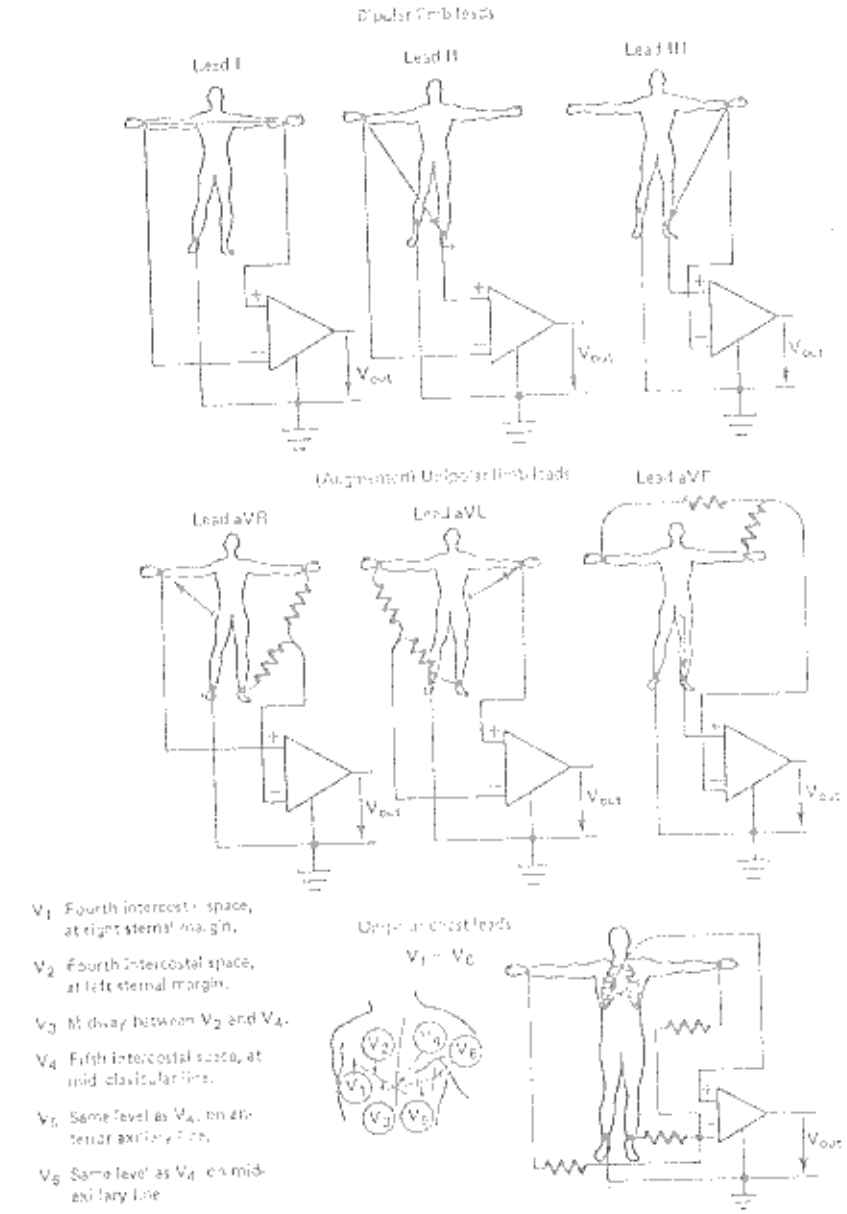

Bipolar limb leads
Lead I
Lead II
Lead III
Augmented unipolar limb leads
Lead aVR
Lead aVL
Lead aVF
V_1 Fourth intercostal space, at right sternal margin.
V_2 Fourth intercostal space, at left sternal margin.
V_3 Midway between V_2 and V_4.
V_4 Fifth intercostal space, at mid-clavicular line.
V_5 Same level as V_4, on anterior axillary line.
V_6 Same level as V_4, on mid-axillary line.
Unipolar chest leads
$V_1 - V_6$

Figure 6.4. ECG lead configurations.

The sides of the triangle represent the lines along which the three projections of the ECG vector are measured. Based on this, Einthoven showed that the instantaneous voltage measured from any one of the three limb lead positions is approximately equal to the algebraic sum of the other two, or

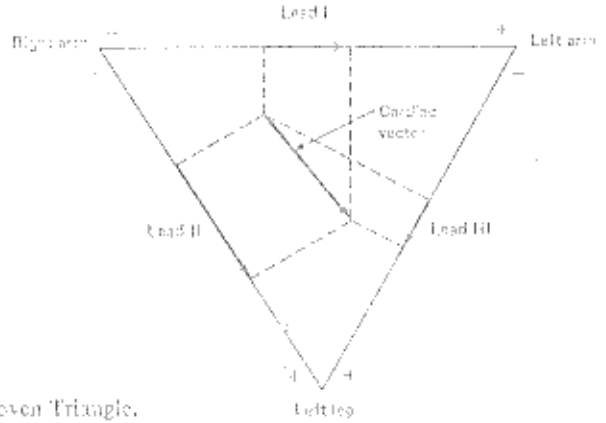


Figure 6.5. The Einthoven Triangle.

that the vector sum of the projections on all three lines is equal to zero. For these statements to actually hold true, the polarity of the lead II measurement must be reversed.

Of the three limb leads, lead II produces the greatest R-wave potential. Thus, when the amplitudes of the three limb leads are measured, the R-wave amplitude of lead II is equal to the sum of the R-wave amplitudes of leads I and III.

The other leads shown in Figure 6.4 are of the *unipolar* type, which was introduced by Wilson in 1944. For unipolar leads, the electrocardiogram is recorded between a single *exploratory electrode* and the *central terminal,* which has a potential corresponding to the center of the body. This central terminal is obtained by connecting the three active limb electrodes together through resistors of equal size. The potential at the connection point of the resistors corresponds to the mean or average of the potentials at the three electrodes. In the *unipolar limb leads,* one of the limb electrodes is used as an exploratory electrode as well as contributing to the central terminal. This double use results in an ECG signal that has a very small amplitude. In *augmented unipolar limb leads,* the limb electrode used as an exploratory electrode is not used for the central terminal, thereby increasing the amplitude of the ECG signal without changing its waveform appreciably. These leads are designated aVR, aVL, and aVF (F as in foot).

For the *unipolar chest leads,* a single chest electrode (exploring electrode) is sequentially placed on each of the six predesignated points on the chest. These chest positions are called the *precordial unipolar* leads and are designated V_1 through V_6. These leads are diagrammed in the lower part of Figure 6.4. All three active limb electrodes are used to obtain the central terminal, while a separate chest electrode is used as an exploratory electrode.

The electrocardiograms recorded from these 12 lead selections are shown in Figure 6.6. It can be seen that the trace from lead selection I or II resembles most closely the idealized waveform of Figure 6.1; some of the other traces are quite different in appearance.

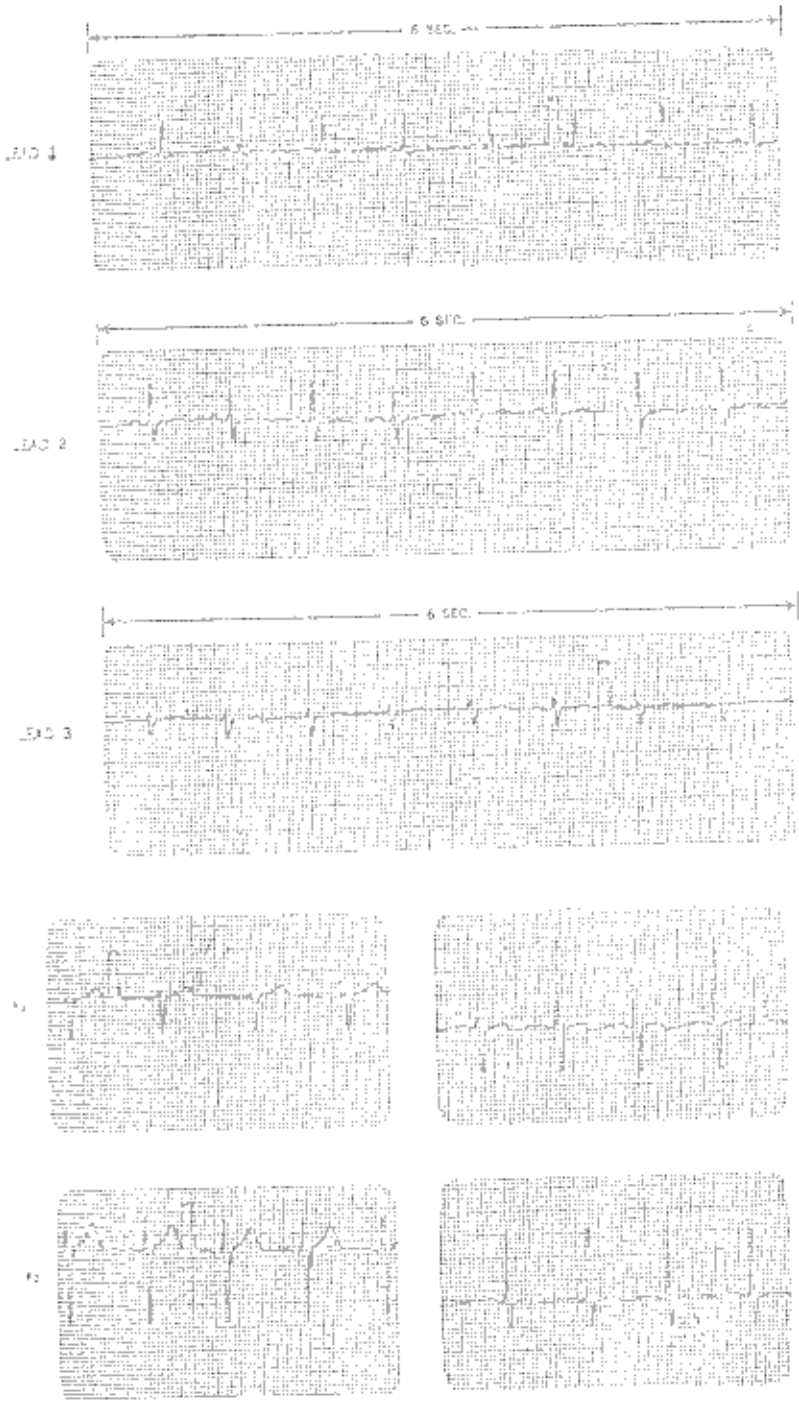


Figure 6.6. Typical patient ECG.

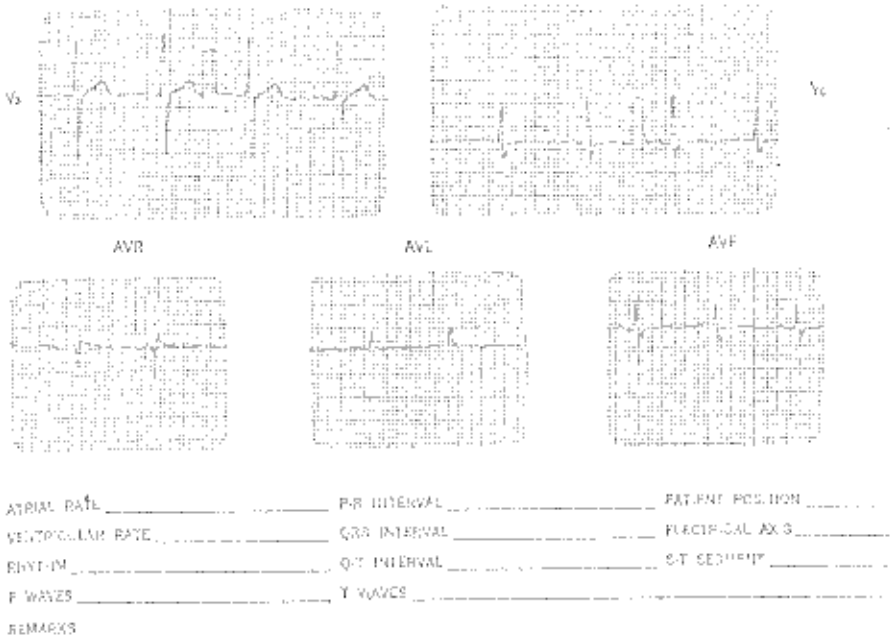


Figure 6.6. Continued

In addition to the lead systems already discussed, there are certain additional lead modifications that are of considerable use in the coronary care unit. The most widely used modification for ongoing ECG monitoring is the *modified chest lead I* (MCL_1) also called the *Marriott lead,* named after its inventor. This lead system simulates the V_1 position with electrode placement as follows: positive electrode, fourth intercostal space, right sternal border; negative electrode just below the outer portion of the left clavicle, with ground just about anywhere, but usually below the right clavicle. The monitor is set on lead I for this bipolar tracing. Recordings obtained in this way are very useful in differentiating left ventricular ectopic rhythms from aberrant right ventricular or supraventricular rhythms. The former situation usually necessitates prompt therapeutic action; the latter is of less clinical significance.

6.1.4. ECG Recorder Principles

The principal parts or building blocks of an ECG recorder are shown in Figure 6.7. Also shown are the controls usually found on ECG recorders; the dashed lines indicate the building block with which each control interacts.

The connecting wires for the patient electrodes originate at the end of a *patient cable,* the other end of which plugs into the ECG recorder. The wires from the electrodes connect to the *lead selector switch,* which also incorporates the resistors necessary for the unipolar leads. A pushbutton

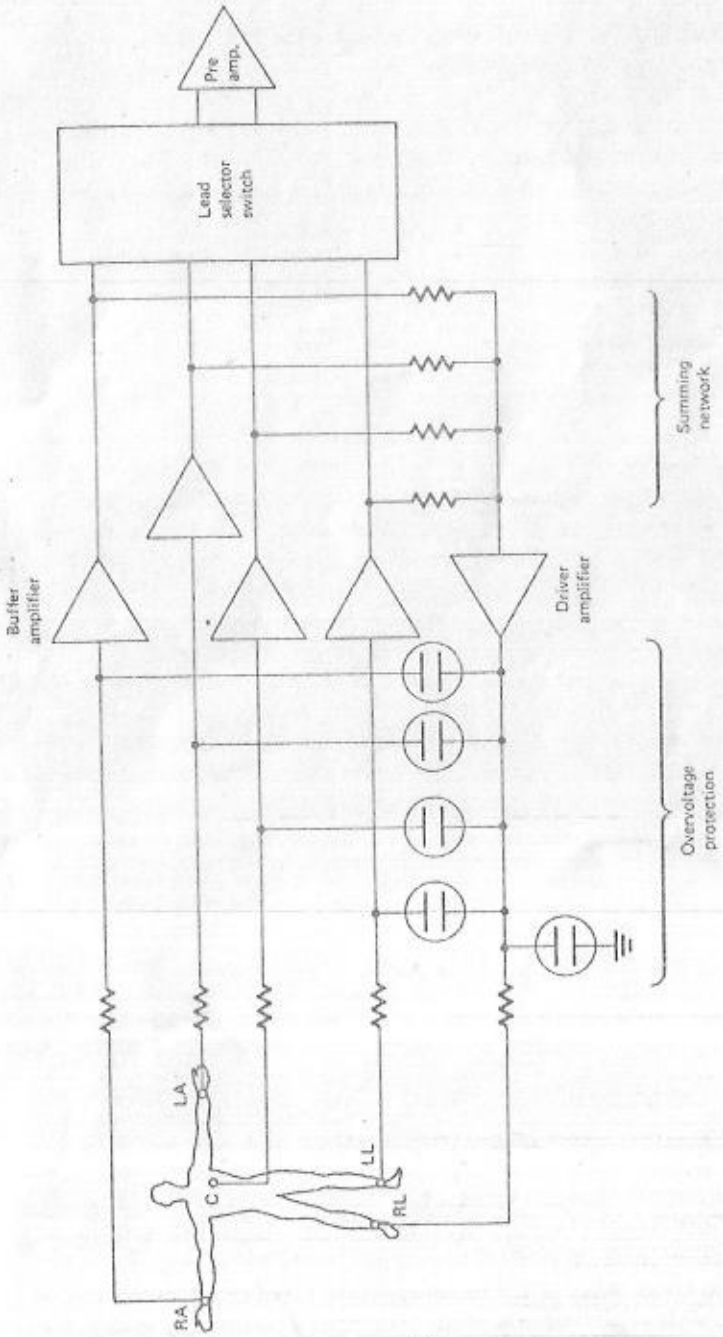


Figure 6.8. Input circuit of modern ECG machine with buffer amplifier; driven right-leg lead andover-voltage protection.

connecting the electrodes to the patient. Modern ECG machines, which have line plugs with grounding pins, do not require such a polarity test.

Although Figure 6.7 shows the principal building blocks of an electrocardiograph, it does not show the circuit details that are found in modern devices of this type. Some of these features are shown in Figure 6.8. To increase the input impedance and thus reduce the effect of variations in electrode impedance, these instruments usually include a *buffer amplifier* for each patient lead. The transistors in these amplifiers are often protected by a network of resistors and neon lamps from overvoltages that may occur when the electrocardiograph is used during surgery in conjunction with high-frequency devices for cutting and coagulation.

A more severe problem is the protection of the electrocardiograph from damage during defibrillation. The voltages that may be encountered in this case can reach several thousand volts. Thus, special measures must be incorporated into the electrocardiograph to prevent burnout of components and provide fast recovery of the trace so as to permit the success of the countershock to be judged.

Some modern devices do not connect the right leg of the patient to the chassis, but utilize a "driven right leg lead." This involves a summing network to obtain the sum of the voltages from all other electrodes and a driving amplifier, the output of which is connected to the right leg of the patient. The effect of this arrangement is to force the reference connection at the right leg of the patient to assume a voltage equal to the sum of the voltages at the other leads. This arrangement increases the common-mode rejection ratio of the overall system and reduces interference. It also has the effect of reducing the current flow in the right leg electrode. Increased concern for the safety aspect of electrical connections to the patient have caused modern ECG designs to abandon the principle of a ground reference altogether and use isolated or floating-input amplifiers, as described in Chapter 16.

6.1.5. Types of ECG Recorders

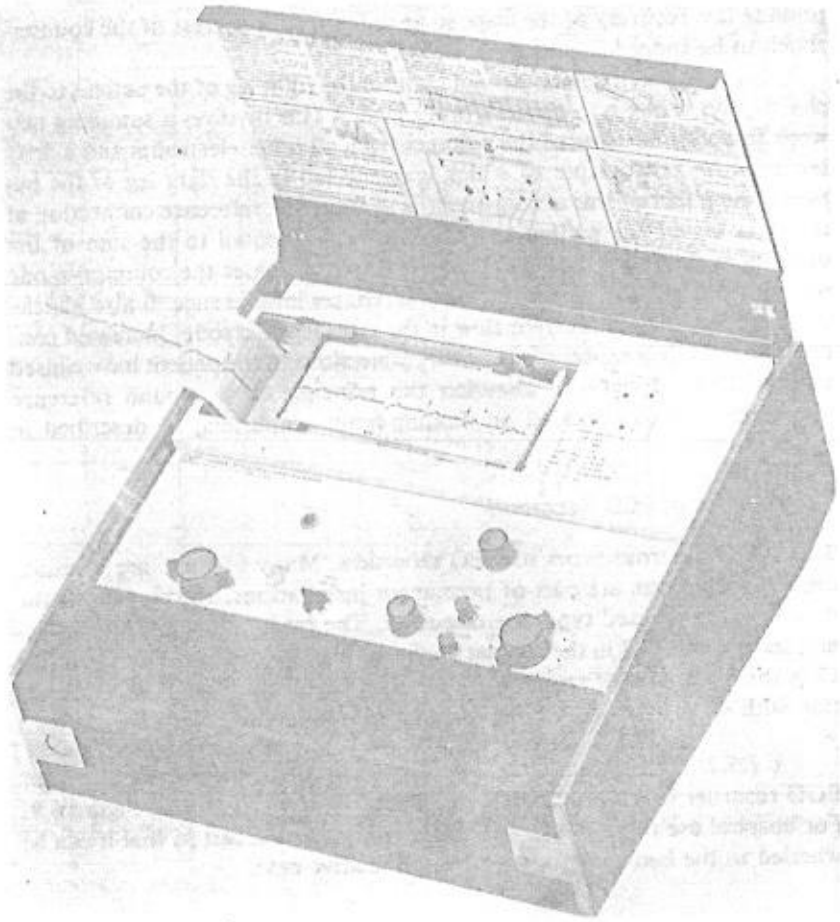
There are numerous types of ECG recorders. Many of these are portable units, while others are part of permanent installations. In this section the most commonly used types are discussed. The reader will find further examples in Chapter 7 in the context of the coronary-care unit and in Chapter 12 in the discussion of emergency care. The use of the computer in connection with electrocardiography is covered in Chapter 15.

6.1.5.1. Single-channel recorders. The most frequently used type of EGG recorder is the portable single-channel unit illustrated in Figure 6.9. For hospital use this recorder is usually mounted on a cart so that it can be wheeled to the bedside of a patient with relative ease.

If the electrocardiogram of a patient is recorded in the 12 standard lead configurations, the resulting paper strip is from 3 to 6 ft long. Even if folded in accordion fashion, the strip is still inconvenient to read and store. Therefore, it is usually cut up, and sections of the recordings from the 12 leads are mounted as shown in Figure 6.6. Because it is easy to mix up the cut sections, the lead for each trace is encoded at the margin of the paper, using the marker pen, during the recording process. The code markers consist of short marks (dots) and long marks (dashes) and look similar to Morse code. No standard code has been established for this purpose, however.

The cut sections of the electrocardiogram can be mounted by inserting them in pockets of a special folder with cutouts to make the trace visible.

Figure 6.9. Single-channel portable ECG machine. (Courtesy of Hewlett-Packard Co., Andover, MA.)



This is the way in which the electrocardiogram in Figure 6.6 was mounted. It should be noted that the recordings from the three limb leads are longer than those from the other lead selections in order to show several QRS complexes; they are called *rhythm strips.* Commercial systems are available to simplify the mounting by die-cutting the paper strip and using mounting cards with adhesive pads. With the automatic three-channel recorders described in the next section, the mounting is greatly simplified.

6.1.5.2. Three-channel recorders. Where large numbers of electrocardiograms are recorded and mounted daily, substantial savings in personnel can be achieved by the use of *automatic three-channel recorders.* These devices not only record three leads simultaneously on a three-channel recorder, but they also switch automatically to the next group of three leads. An electrocardiogram with the 12 standard leads, therefore, can be recorded automatically as a sequence of four groups of three traces. The time required for the actual recording is only about 10 seconds. The groups of leads recorded and the time at which the switching occurs are automatically identified by code markings at the margin of the recording paper. At the end of the recording, standardization pulses are inserted in all three recording channels. Although the actual recording time is reduced substantially compared to single-channel recorders, more time is required to apply the electrodes to the patient because separate electrodes must necessarily be used for each chest position. The mounting of the electrocardiogram, however, is simplified substantially, for no cutting or mounting of the individual lead selections is required. A modern recorder of this type is shown in Figure 6.10.

6.1.5.3. Vector electrocardiographs (vectorcardiographs). As noted in Section 6.1.3, the voltage generated by the activity of the heart can be described as a vector whose magnitude and spatial orientation change with time. In the type of electrocardiography described thus far, only the magnitude of the voltage is recorded. *Vectorcardiography,* on the other hand, presents an image of both the magnitude and the spatial orientation of the heart vector. The heart vector, however, is a three-dimensional variable, and three "views" or projections on orthogonal planes are necessary to describe the variable fully in two-dimensional figures. Special lead placement systems must be used to pick up the ECG signals for vector electrocardiograms, the *Frank system* being the one most frequently employed. The vectorcardiogram is usually displayed on a cathode-ray tube, similar to those used for patient monitors. Each QRS complex is displayed as a sequence of "loops" on this screen, which is then photographed with a Polaroid camera. Vectorcardiographs that use computer techniques to slow down the ECG signals and allow the recording of the vectorcardiogram with a mechanical X-Y recorder are also available.

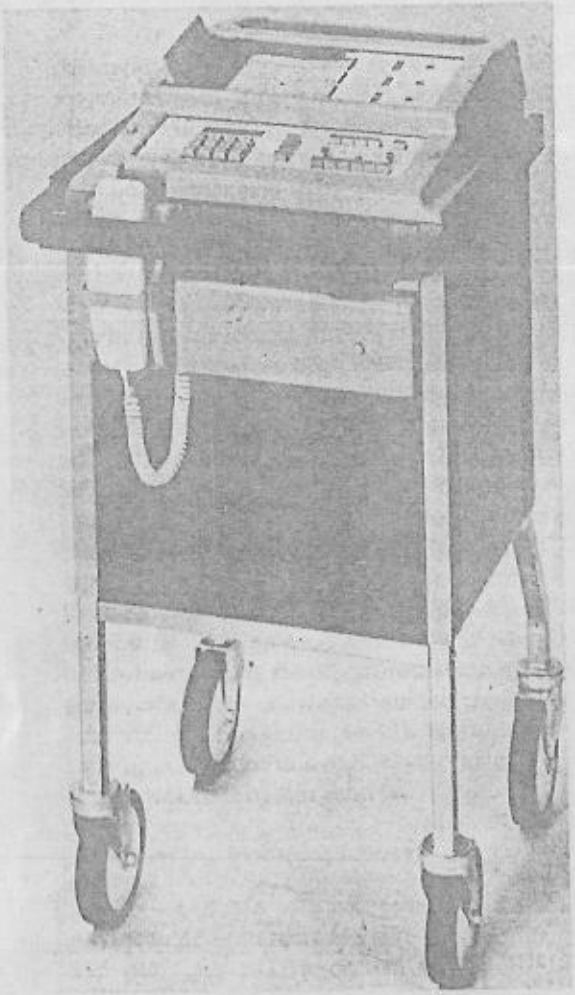


Figure 6.10. Automatic three-channel ECG recorder with keyboard for entering patient data and telephone coupler to send ECG and data to a remote computer via telephone lines. (Courtesy of Hewlett-Packard Co., MA.)

6.1.5.4. Electrocardiograph systems for stress testing. Coronary insufficiency frequently does not manifest itself in the electrocardiogram if the recording is taken during rest. In the *Masters test* or *two-step exercise test,* a physiological stress is imposed on the cardiovascular system by letting the patient repeatedly walk up and down a special pair of 9-inch high steps prior to recording his electrocardiogram. Based on the same principle is the *exercise stress test,* in which the patient walks at a specified speed on a treadmill whose inclination can be changed. While the Masters test is normally conducted using a regular single-channel electrocardiograph, special systems are available for the exercise stress test. These systems, however, are usually made up of a number of individual instruments which are described in this book.

An exercise stress test system typically consists of the following parts:

1. A treadmill which may incorporate an automatic programmer to change the speed and inclination in order to apply a specific physiological stress.
2. An ECG radiotelemetry system to allow recording of the ECG without artifacts while the patient is on the treadmill.
3. An ECG monitor with a cathode-ray-tube display and heart rate meter.
4. An ECG recorder.
5. An automatic or semiautomatic sphygmomanometer for the indirect measurement of blood pressure.

Because the exercise stress test involves a certain risk for patients with known or suspected cardiac disorders, a dc defibrillator is usually kept available while the test is performed.

6.1.5.5. Electrocardiographs for computer processing. The automatic analysis of electrocardiograms by computers is used increasingly (see Chapter 15). This technique requires that the ECG signal from the standard leads be transmitted sequentially to the computer by some suitable means, together with additional information on the patient. The automatic three-channel recorders can frequently be adapted for this purpose. The ECG signals can either be recorded on a tape for later computer entry or can be directly transmitted to the computer through special lines or regular telephone lines using a special acoustical coupler (see Chapter 12). Information regarding the patient is entered with thumbwheel switches or from a keyboard and is transmitted along with the ECG signal. During the transmission of the signal, the electrocardiogram is simultaneously recorded to verify that the transmitted signals are free of artifacts.

6.1.5.6. Continuous ECG recording (Holter recording). Because a normal electrocardiogram represents only a brief sample of cardiac activity, arrhythmias which occur intermittently or only under certain conditions, such as emotional stress, are frequently missed. The technique of continuous ECG recording, which was introduced by Norman Holter, makes it possible to capture these kinds of arrhythmias. To obtain a continuous ECG, the electrocardiogram of a patient is recorded during his normal daily activity by means of a special magnetic tape recorder. The smallest device of this type can actually be worn in a shirt pocket and allows recordings of the ECG for four hours. Other recorders, about the size of a camera case, are worn over the shoulder and can record the electrocardiogram for up to 24 hours. The recorded tape is analyzed using a special scanning device which plays back the tape at a higher speed than that used for recording. By this method

a 24-hour tape can be reviewed in as little as 12 minutes. During the playback, the beat-to-beat interval of the electrocardiogram is displayed on a cathode-ray tube as a picket-fence-like pattern in which arrhythmia episodes are clearly visible. Once such an episode has been discovered, the tape is backed up and slowed down to obtain a normal electrocardiogram strip for the time interval during which the arrhythmias occurred. A special time clock is synchronized by the tape drive to correlate the onset of the episode with the activity of the patient.

6.2. MEASUREMENT OF BLOOD PRESSURE

As one of the physiological variables that can be quite readily measured, blood pressure is considered a good indicator of the status of the cardiovascular system. A history of blood pressure measurements has saved many a person from an untimely death by providing warnings of dangerously high blood pressure (*hypertension*) in time to provide treatment.

In routine clinical tests, blood pressure is usually measured by means of an indirect method using a *sphygmomanometer* (from the Greek word, *sphygmos*, meaning pulse). This method is easy to use and can be automated. It has, however, certain disadvantages in that it does not provide a continuous recording of pressure variations and its practical repetition rate is limited. Furthermore, only systolic and diastolic arterial pressure readings can be obtained, with no indication of the details of the pressure waveform. The indirect method is also somewhat subjective, and often fails when the blood pressure is very low (as would be the case when a patient is in shock).

Methods for direct blood pressure measurement, on the other hand, do provide a continuous readout or recording of the blood pressure waveform and are considerably more accurate than the indirect method. They require, however, that a blood vessel be punctured in order to introduce the sensor. This limits their use to those cases in which the condition of the patient warrants invasion of the vascular system.

This section is divided into three parts. First, indirect or noninvasive methods are discussed. Since there has been much progress in the automating of indirect techniques, automated methods are covered in a separate section. Finally, direct or invasive blood pressure measurements are discussed.

6.2.1. Indirect Measurements.

As stated earlier, the familiar indirect method of measuring blood pressure involves the use of a sphygmomanometer and a stethoscope. The

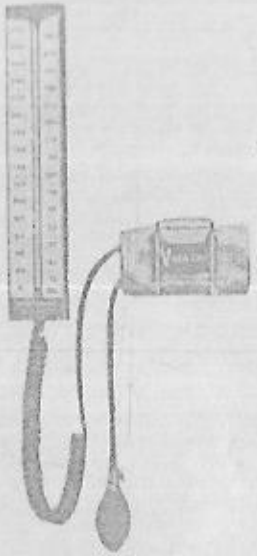


Figure 6.11. Wall-mounted sphygmomanometer. (Courtesy of W.A. Baum, Inc., Copiague, NY.)

sphygmomanometer consists of an inflatable pressure cuff and a mercury or aneroid manometer to measure the pressure in the cuff. The cuff consists of a rubber bladder inside an inelastic fabric covering that can be wrapped around the upper arm and fastened with either hooks or a Velcro fastener. The cuff is normally inflated manually with a rubber bulb and deflated slowly through a needle valve. The stethoscope is described in detail in Section 6.5. A wallmounted sphygmomanometer is shown in Figure 6.11. These devices are also manufactured as portable units.

The sphygmomanometer works on the principle that when the cuff is placed on the upper arm and inflated, arterial blood can flow past the cuff only when the arterial pressure exceeds the pressure in the cuff. Furthermore, when the cuff is inflated to a pressure that only partially occludes the brachial artery, turbulence is generated in the blood as it spurts through the tiny arterial opening during each systole. The sounds generated by this turbulence, *Korotkoff sounds*, can be heard through a stethoscope placed over the artery downstream from the cuff.

To obtain a blood pressure measurement with a sphygmomanometer and a stethoscope, the pressure cuff on the upper arm is first inflated to a pressure well above systolic pressure. At this point no sounds can be heard through the stethoscope, which is placed over the brachial artery, for that artery has been collapsed by the pressure of the cuff. The pressure in the cuff is then gradually reduced. As soon as cuff pressure falls below systolic pressure, small amounts of blood spurt past the cuff and Korotkoff sounds begin to be heard through the stethoscope. The pressure of the cuff that is indicated on the manometer when the first Korotkoff sound is heard is recorded as the systolic blood pressure.

Figure 6.12. Measurement of blood pressure using sphygmomanometer. (Courtesy of W.A. Baum, Inc., Copiague, NY.)

As the pressure in the cuff continues to drop, the Korotkoff sounds continue until the cuff pressure is no longer sufficient to occlude the vessel during any part of the cycle. Below this pressure the Korotkoff sounds disappear, marking the value of the diastolic pressure.

This familiar method of locating the systolic and diastolic pressure values by listening to the Korotkoff sounds is called the *auscultatory method* of sphygmomanometry. An alternative method, called the *palpatory method,* is similar except that the physician identifies the flow of blood in the artery by feeling the pulse of the patient downstream from the cuff instead of listening for the Korotkoff sounds. Although systolic pressure can easily be measured by the palpatory method, diastolic pressure is much more difficult to identify. For this reason, the auscultatory method is more commonly used. Figure 6.12 shows a blood pressure measurement using the auscultatory method.

6.2.2. Automated Indirect Methods

Because of the trauma imposed by direct measurement of blood pressure (described below) and the lack of a more suitable method for indirect measurement, attempts have been made to automate the indirect procedure. As a result, a number of automatic and semiautomatic systems have been developed. Most devices are of a type that utilizes a pressure transducer

connected to the sphygmomanometer cuff, a microphone placed beneath the cuff (over the artery), and a standard physiological recording system on which cuff pressure and the Korotkoff sounds are recorded. The basic procedure essentially parallels the manual method. The pressure cuff is automatically inflated to about 220 mm Hg and allowed to deflate slowly. The microphone picks up the Korotkoff sounds from the artery near the surface, just below the compression cuff. One type of instrument either superimposes the signal of the Korotkoff sounds on the voltage recording representing the falling cuff pressure or records the two separately. The pressure reading at the time of the first sound represents the systolic pressure; the diastolic pressure is the point on the falling pressure curve where the signal representing that last sound is seen. This instrument is actually only semiautomatic because the recording thus obtained must still be interpreted by the observer. False indications—caused, for instance, by motion artifacts—can often be observed on the recording. Fully automated devices use some type of signal-detecting circuit to determine the occurrence of the first and last Korotkoff sounds and retain and display the cuff pressure reading for these points, either electronically or with mercury manometers that are cut off by solenoid valves. One of the recent innovations of these techniques is the "do-it-yourself" blood pressure machine to be found in some supermarkets. These devices, by necessity, are more susceptible to false indications caused by artifacts.

Many of the commercially available automatic blood pressure meters work well when demonstrated on a quiet, healthy subject but fail when used to measure blood pressure during activity or when used on patients in circulatory shock. Methods other than those utilizing the Korotkoff sounds have been tried in detecting the blood pulse distal to the occlusion cuff. Among them is impedance plethysmography (see Section 6.4), which indicated directly the pulsating blood flow in the artery, and ultrasonic Doppler methods, which measure the motions of the arterial walls. An early example of an automatic blood pressure meter is the *programmed electrosphygmomanometer PE-300,* illustrated in Figures 6.13 and 6.14 in block diagram and pictorial form. This instrument is designed for use in conjunction with an occluding cuff, microphone, or pulse transducer, and a recorder for the automatic measurement of indirect systolic and diastolic blood pressures from humans and many animal subjects.

The PE-300 incorporates a transducer-preamplifier that provides two output signals, a voltage proportional to the cuff pressure, and the amplified Korotkoff sounds or pulses. These signals can be monitored individually or with the sounds or pulses superimposed on the calibrated cuff-recorder. The combined signal can be recorded on a graphic pen recorder.

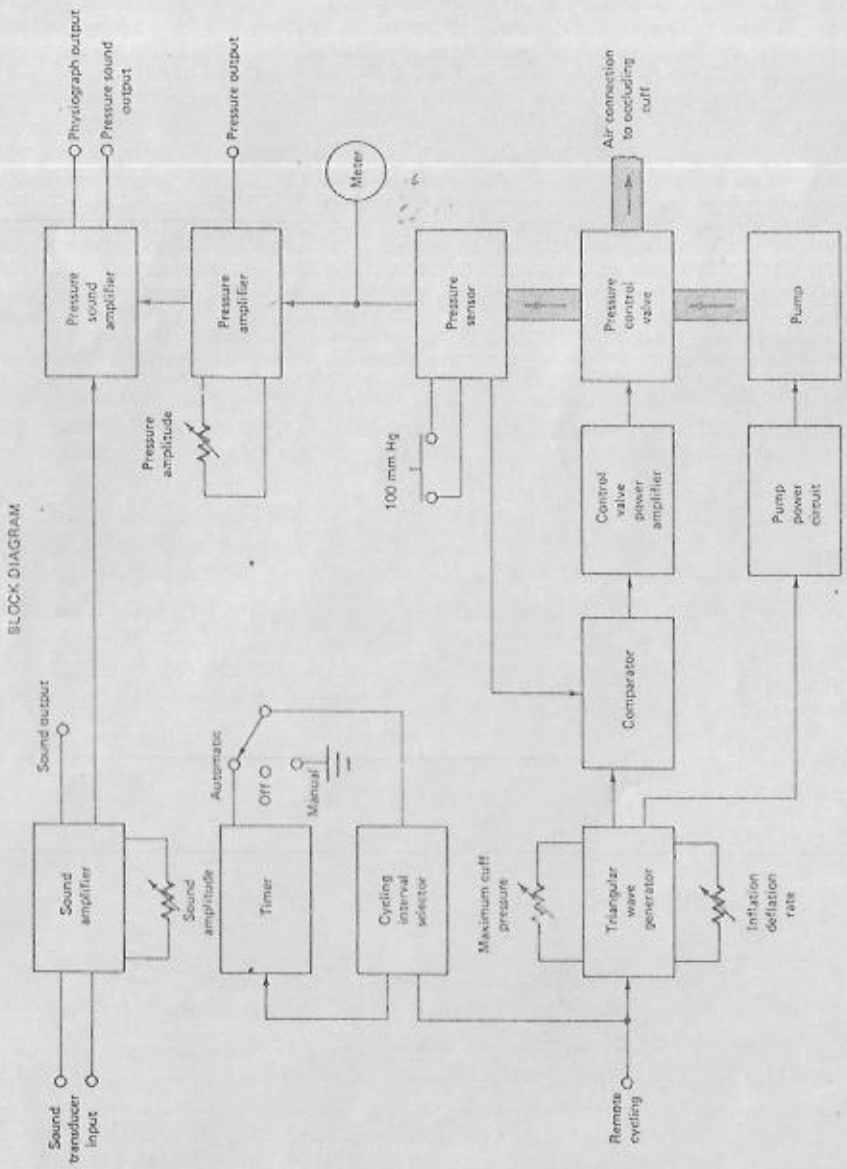


Figure 6.13. Programmed electrosphygmomanometer (block diagram). (Courtesy of Narco BioSystems, Inc., Houston, TX.)

Figure 6.14. Electrosphygmomanometer. (Courtesy of Narco BioSystems, Inc., Houston, TX.)

The self-contained cuff inflation system can be programmed to inflate and deflate an occluding cuff at various rates and time intervals. Equal and linear rates of cuff inflation and deflation permit two blood pressure determinations per cycle. The PE-300 can be programmed for repeat cycles at adjustable time intervals for monitoring of blood pressure over long periods of time. Single cycles may be initiated by pressing a panel-mounted switch. Provision is also made for remote control via external contact closure. The maximum cuff pressure is adjustable, and the front-panel meter gives a continuous visual display of the cuff pressure.

The electronic sphygmomanometer shown in Figure 6.15 is an example of a device incorporating a large gage for easy reading. The scale is 6 in. (over 15 cm) in diameter. In this example the cuff has to be inflated manually and so this instrument is sometimes called "semiautomatic." The Korotkoff sounds are automatically detected by microphones and a light flashes at systolic pressure, which stops at the diastolic value.

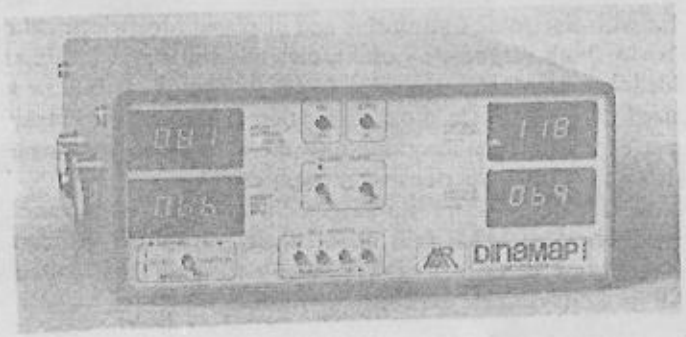
While the clinical diagnostic value of systolic and diastolic blood pressure has been clearly established, the role of *mean arterial pressure* (MAP) as an indication of blood pressure trend has become more widely accepted with the expanded use of direct pressure monitoring using arterial cannulae with electronic transducers and displays. Most electrical monitors now provide both diagnostic systolic/diastolic waveform information and the added option of a single-value MAP indication. It is now generally recognized that MAP is a direct indication of the pressure available for tissue perfusion, and that a continuously increasing or decreasing MAP can ultimately result in a hypertensive or hypotensive crisis.

Mean arterial pressure is a weighted average of systolic and diastolic pressure. Generally, MAP falls about one-third of the way between the diastolic low and the systolic peak. A simple formula for calculating MAP is: MAP = 1/3(systolic - diastolic) + diastolic.

Figure 6.15. Electronic sphygmomanometer. (Courtesy of Applied Science Laboratories, Waltham, MA.)

In the Dinamap illustrated in Figure 6.16, the mean arterial blood pressure is determined by the oscillometric method. The pressure pulsations or oscillations introduced within a cuff bladder are sensed by a solid-state pressure transducer located within the enclosure. As the air pressure in the cuff is decreased, pressure oscillations increase in amplitude and reach a maximum as the cuff pressure passes through the pressure equal to the mean arterial blood pressure. This phenomenon can also be observed as it produces pulsations in the mercury column of a conventional sphygmomanometer or the needle oscillations in an aneroid sphygmomanometer.

Figure 6.16. Dinamap for automatic detection of mean arterial blood pressure. (Courtesy of Applied Medical Research, Tampa, FL.)



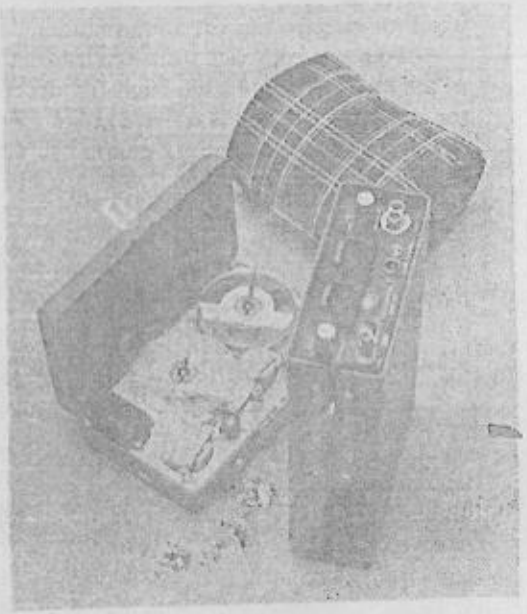
Since the indications associated with the indirect measurement of mean arterial pressure pass through a maximum at the point of interest, as opposed to the phenomenon for the indirect determination of systolic and diastolic pressure, which is at a minimum at the point of interest, the Dinamap can determine arterial pressures over a wide range of physiological states. It works effectively on most patients on whom systolic and diastolic indirect pressures are impossible to determine or are subject to such variabilities as to be of questionable clinical value.

An internal microprocessor rejects most artifacts caused by patient movement or external interference. Under program control, all normal operating variables, such as inflation pressures, deflation rates, alarm limits, and abnormal operation alarms, are automatically determined and controlled without operator adjustments. Adjustable alarm limits and measurement cycle times are simply set via front panel switches by the user as required for varying clinical conditions.

Because, in most clinical situations, systolic and diastolic pressures correlate with each other and MAP is determined from the systolic/diastolic pair, critical patient arterial pressure trends can be monitored by observation of MAP. Adjustable alarm limits, as required for a given clinical situation, can then alert staff of possible patient problems. These units can be used in operating rooms, recovery rooms, and intensive-care units.

The ability to measure blood pressure automatically with portable equipment makes it possible to take measurements while the patient is pursuing his or her normal activities. A system that does this is shown in Figure

Figure 6.17. Ambulatory automatic blood pressure monitor. (Courtesy of Del Mar Avionics, Irvine, CA.)



6.17. This device is a 24-hour automatic noninvasive blood pressure and Holter (see Section 6.1.5.6) monitoring system with trend writeouts. The three major components of the ambulatory system are shown in the photograph. The Pressurometer II Model 1977 at the front is the blood pressure measuring device. The recording unit is the Model 446A Electrocardiocorder, which records blood pressure and ECG on a 24-hour basis. The patient wears the units with belts and straps. The cuff is attached comfortably and securely on the left arm.

The Model 1977 utilizes a standard pneumatic cuff. A transducer for the detection of the Korotkoff sounds is held in place with an adhesive disk. Cuff pressure is applied automatically and the patient's systolic and diastolic pressures are measured in excess of 100 preprogrammed intervals in a 24-hour period. The preselected intervals can be overridden by means of manual switching and the unit can be operated manually for checking and calibration. When used with the 446A recorder, the Korotkoff sounds are gated by ECG R-wave signals. The recordings can be fed into a companion trend computer, which gives a digital readout of the data on the recording. The entire system is powered by a portable rechargeable nickelcadmium battery pack which permits up to 26 hours of recording. The computer is a plug-in module and chart-paper documentation is also available.

An important feature of the system is that it can be used with an Electrocardioscanner, which scans all the data at a rate 120 times as fast as they were recorded (the entire 24-hour record can be scanned in 12 minutes) quantitating ectopic beats and total heart beats and printing hourly trends on heart rate and other quantities. It can rewind itself and will search out operator-selected patient abnormalities.

Another approach utilizes ultrasound to measure the pulsatile motion of the brachial artery wall. High-frequency sound energy is transmitted into the patient's arm and is reflected back from the arterial walls. By means of the Doppler effect (see explanation in Chapter 9), the movement of the arterial walls can be detected as they snap open and closed with each pulsation of blood.

An advantage of this type of instrument is that results closer to direct measurements (Section 6.2.3) can be obtained. Also, it can be used for patients under shock and in intensive care units, when direct measurement would not be suitable for the patient. Because vessel-wall movement is sensed, blood flow is not a requirement for measurement.

One such instrument is the Arteriosonde, which has a cloth cuff and air bag with an electric air pump to supply the pressure. The pump can be regulated by a front-panel control. The cuff is placed on the arm in the same fashion as the sphygmomanometer except that there is a transducer array under the cuff. These transducers are arranged as alternate transmitters and receivers. The motion of the artery produces a *Doppler shift*, which identifies the instant the artery is opened and closed with each beat between systolic and diastolic pressure.

The first signal is used to stop the mercury fall in the first of two manometers to indicate systolic pressure. The second manometer is stopped at the point of disappearance of the pulses to indicate diastolic pressure. Hence, both readings can be noted simultaneously.

6.2.3. Direct Measurements

In 1728, Hales inserted a glass tube into the artery of a horse and crudely measured arterial pressure. Poiseuille substituted a mercury manometer for the piezometer tube of Hales, and Ludwig added a float and devised the *kymograph*, which allowed continuous, permanent recording of the blood pressure. It is only quite recently that electronic systems using strain gages as transducers have replaced the kymograph.

Regardless of the electrical or physical principles involved, direct measurement of blood pressure is usually obtained by one of three methods:

1. Percutaneous insertion.
2. Catheterization (vessel cutdown).
3. Implantation of a transducer in a vessel or in the heart.

Other methods, such as clamping a transducer on the intact artery, have also been used, but they are not common.

Figure 6.18 should give a general idea of both methods. Typically, for percutaneous insertion, a local anesthetic is injected near the site of invasion. The vessel is occluded and a hollow needle is inserted at a slight angle toward the vessel. When the needle is in place, a catheter is fed through the hollow needle, usually with some sort of a guide. When the catheter is securely in place in the vessel, the needle and guide are withdrawn. For some measurements, a type of needle attached to an airtight tube is used, so that the needle can be left in the vessel and the blood pressure sensed directly by attaching a transducer to the tube. Other types have the transducer built into the tip of the catheter. This latter type is used in both percutaneous and full catheterization models.

Catheterization was first developed in the late 1940s and has become a major diagnostic technique for analyzing the heart and other components of the cardiovascular system. Apart from obtaining blood pressures in the heart chambers and great vessels, this technique is also used to obtain blood samples from the heart for oxygen-content analysis and to detect the location of abnormal blood flow pathways. Also, catheters are used for investigations with injection of radiopaque dyes for X-ray studies, colored dyes for indicator dilution studies, and of vasoactive drugs directly into the heart and certain vessels. Essentially, a *catheter* is a long tube that is in-

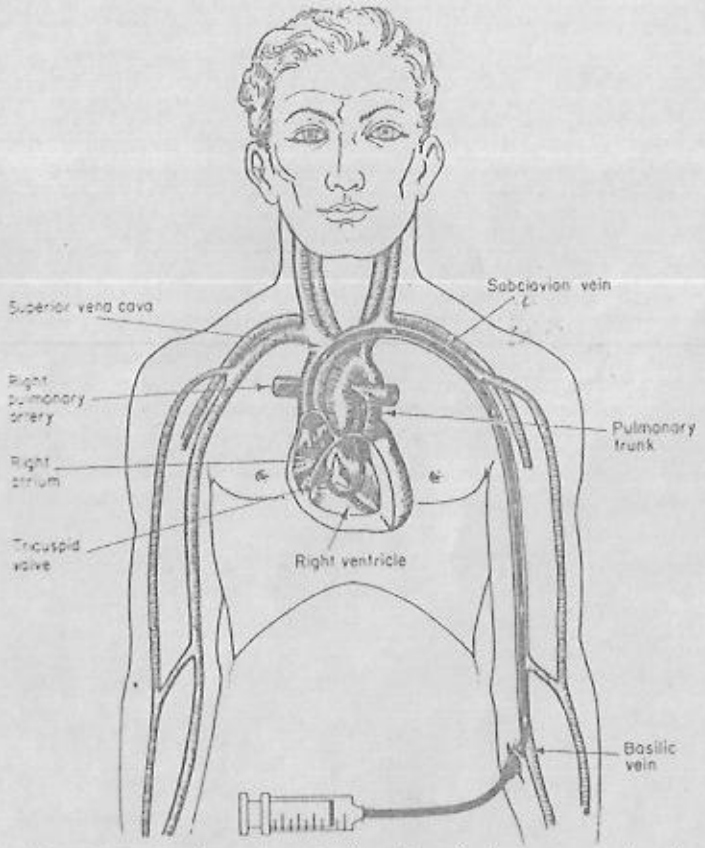


Figure 6.18. Cardiac catheterization. The tube is shown entering the basilic vein in this case. (From W.F. Evans, *Anatomy and Physiology, The Basic Principles,* Englewood Cliffs, NJ., Prentice-Hall, Inc., 1971, by permission.)

troduced into the heart or a major vessel by way of a superficial vein or artery. The sterile catheter is designed for easy travel through the vessels.

Measurement of blood pressure with a catheter can be achieved in two ways. The first is to introduce a sterile saline solution into the catheter so that the fluid pressure is transmitted to a transducer outside the body (extracorporeal). A complete fluid pressure system is set up with provisions for checking against atmospheric pressure and for establishing a reference point. The frequency response of this system is a combination of the frequency response of the transducer and the fluid column in the catheter. In the second method, pressure measurements are obtained at the source. Here, the transducer is introduced into the catheter and pushed to the point at which the pressure is to be measured, or the transducer is mounted at the tip of the catheter. This device is called a *catheter-tip blood pressure*

transducer. For mounting at the end of a catheter, one manufacturer uses an unbonded resistance strain gage in the transducer, whereas another uses a variable inductance transducer (see Chapter 2). Each will be discussed later.

Implantation techniques involve major surgery and thus are normally employed only in research experiments. They have the advantage of keeping the transducer fixed in place in the appropriate vessel for long periods of time. The type of transducer employed in that procedure is also described later in this section.

Transducers can be categorized by the type of circuit element used to sense the pressure variations, such as capacitive, inductive, and resistive. Since the resistive types are most frequently used, the other two types are discussed only briefly.

In the *capacitance manometer,* a change in the distance between the plates of a capacitor changes its capacitance. In a typical application, one of the plates is a metal membrane separated from a fixed plate by some one-thousandth of an inch of air. Changes in pressure that change the distance between the plates thereby change the capacitance. If this element is contained in a high-frequency resonant circuit, the changes in capacitance vary the frequency of the resonant circuit to produce a form of frequency modulation. With suitable circuitry, blood pressure information can be obtained and recorded as a function of time.

An advantage of this type of transducer is that its total contour can be long and thin so that it can be easily introduced into the bloodstream without deforming the contour of the recorded pressure waveform. Because of the stiff structure and the small movement of the membrane when pressure is applied, the volume displacement is extremely small (in the region of 10^{-4} cm^3/100 mm Hg of applied pressure).

Disadvantages of this type of transducer are instability and a proneness to variations with small changes in temperature. Also, lead wires introduce errors in the capacitance, and this type of transducer is more difficult to use than resistance types.

A number of different devices use inductance effects. They measure the distortion of a membrane exposed to the blood pressure. In some of these types, two coils are used—a primary and secondary. When a spring-loaded core that couples the coils together magnetically is moved back and forth, the voltage induced into the secondary changes in proportion to the pressure applied.

A better-known method employs a *differential transformer,* described in detail in Chapter 2. In this device two secondary coils are wound oppositely and connected in series. If the spring-loaded core is symmetrically positioned, the induced voltage across one secondary coil opposes the voltage of the other. Movement of the core changes this symmetry, and the

result is a signal developed across the combined secondary coils. The core can be spring-loaded to accept pressure from one side, or it can accept pressure from both sides simultaneously, thus measuring the difference of pressure between two different points.

The *physiological resistance transducer* is a direct adaptation of the strain gages used in industry for many years. The principle of a strain gage is that if a very fine wire is stretched, its resistance increases. (A detailed discussion of strain gages is given in Chapter 2.) If voltage is applied to the resistance, the resulting current changes with the resistance variations according to Ohm's law. Thus, the forces responsible for the strain can be recorded as a function of current. The method by which the blood pressure produces the strain is discussed in Section 6.2.4.

To obtain the degree of sensitivity required for blood pressure transducers, two or four strain gages are mounted on a diaphragm or membrane, and these resistances are connected to form a bridge circuit. Figure 6.19 shows such a circuit configuration.

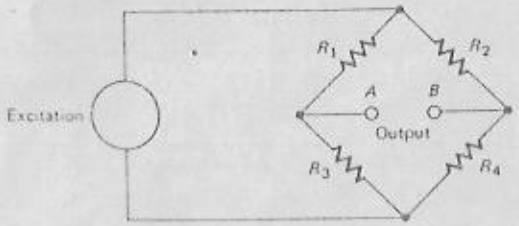


Figure 6.19. Resistance strain-gage bridge.

In general, the four resistances are initially about equal when no pressure or strain is applied. The gages are attached to the pressure diaphragm in such a way that as the pressure increases, two of them stretch while the other two contract. An excitation voltage is applied as shown. When pressure changes unbalance the bridge a voltage appears between terminals *A* and *B* proportional to the pressure. Excitation can be either direct current or alternating current, depending on the application.

Resistance-wire strain gages can be *bonded* or *unbonded* (see Chapter 2). In the bonded type, the gage is "bonded" to the diaphragm and stretches or contracts with bending. The unbonded type consists of two pairs of wires, coiled and assembled in such a way that displacement of a membrane connected to them causes one pair to stretch and the other to relax. The two pairs of wires are not bonded to the diaphragm material but are attached only by retaining lugs. Because the wires are very thin, it is possible to obtain relatively large signals from the bridge with small movement of the diaphragm.

Development of semiconductors that change their resistance in much

the same manner as wire gages has led to the *bonded silicon element* bridge. Only small displacements (on the order of a few micro meters) of the pressure-sensing diaphragm, are needed for sizable changes of output voltage with low-voltage excitation. For example, with 10 V excitation, a range of 300 mm Hg is obtained with a 3-μm deflection, producing a 30-mV signal.

Semiconductor strain-gage bridges are often temperature-sensitive, however, and have to be calibrated for baseline and true zero. Therefore, it is usually necessary to incorporate external resistors and potentiometers to balance the bridge initially, as well as for periodic correction.

In Chapter 2 the gage factor for a strain gage is defined as the amount of resistance change produced by a given change in length. Wire strain gages have gage factors on the order of 2 to 4, whereas semiconductor strain gages have gage factors ranging from 50 to 200. For silicon, the gage factor is typically 120. The use of semiconductors is restricted to those configurations that lend themselves to this technique.

When strain gages are incorporated in pressure transducers, the sensitivity of the transducer is expressed, not as a gage factor, but as a voltage change that results from a given pressure change. For example, the sensitivity of a pressure transducer can be given in microvolts per (applied) volt per millimeter of mercury.

6.2.4. Specific Direct Measurement Techniques

In Section 6.2.3 methods of direct blood pressure were classified in two ways, first by the clinical method by which the measuring device was coupled to the patient and, second, by the electrical principle involved. In the following discussion, the first category is expanded, with the electrical principles involved being used as subcategories where necessary. The four categories are as follows:

1. A catheterization method involving the sensing of blood pressure through a liquid column. In this method the transducer is external to the body, and the blood pressure is transmitted through a saline solution column in a catheter to this transducer. This method can use either an unbonded resistance strain gage to sense the pressure or a linear variable differential transformer. Externally, these two devices are quite similar in appearance.
2. A catheterization method involving the placement of the transducer through a catheter at the actual site of measurement in the bloodstream (e.g., to the aorta), or by mounting the transducer on the tip of the catheter.

3. Percutaneous methods in which the blood pressure is sensed in the vessel just under the skin by the use of a needle or catheter.
4. Implantation techniques in which the transducer is more permanently placed in the blood vessel or the heart by surgical methods.

The most important aspects of these methods are discussed separately.

6.2.4.1. Liquid-column methods. A typical liquid-column blood pressure transducer, the Gould Statham P 23 ID, is illustrated in Figure 6.20. Figure 6.21 is a cutaway drawing to show the interior construction and the isolation features of the same transducer, which is considered a standard size in hospital practice. The heart of the P 23 transducer is the unbonded strain gage, which is connected in a standard Wheatstone bridge configuration. The metal sensing diaphragm can be seen on the left side. It is a precision-made part that must deflect predictably with a given fluid pressure. When the diaphragm is deflected downward by the pressure of the liquid being measured, the tension on two of the bridge wires is relaxed and the tension on the other two wires is tightened, changing the resistance of the gage. For negative pressures, the opposite wires are stretched and relaxed

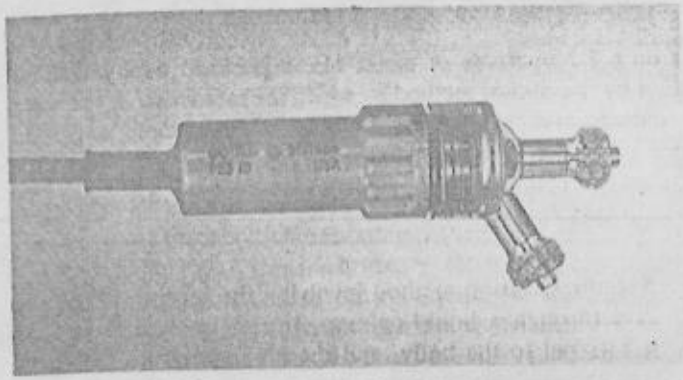


Figure 6.20. Fluid-column blood pressure transducer. (Courtesy of Gould, Inc., Measurement Systems Division, Oxnard, CA.)

The transducer is connected through the cable to an instrument which contains zero-balance and range controls, amplifier circuits, and a readout. The shielded cable is attached to the case through a liquid-tight seal that permits immersion of the transducer for cleaning. The transducer case is vented through the cable so that measurements are always referenced to atmospheric pressure.

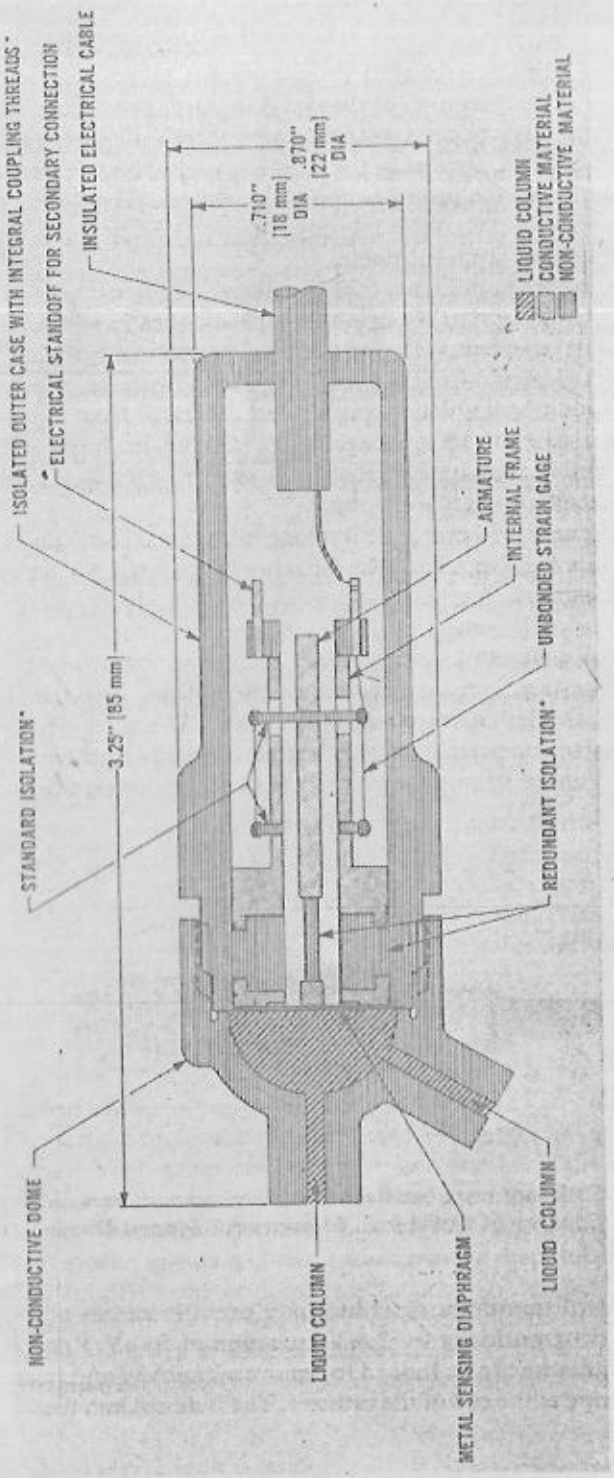


Figure 6.21. Interior construction of P23ID Blood Pressure Transducer. (Courtesy of Gould Inc., Measurement Systems Division, Oxnard, CA.)

The dome is the reservoir for the liquid that transmits blood pressure to the diaphragm. It is made of transparent plastic to facilitate the detection and removal of bubbles, since even the most minute bubble can degrade the frequency response of the pressure-monitoring system. The dome is fitted with two ports. One port is coupled through tubing to the cannula; the other is used for venting air from the dome.

It should be noted in Figure 6.21 that there are three modes of isolation: (1) external isolation of the case with a plastic sheath, which provides protection from extraneous voltages; (2) standard internal isolation of the sensing (bridge) elements from the inside of the transducer case and the frame; and (3) additional isolation (internal) of the frame from the case and the diaphragm in case of wire breakage. Thus, isolation of the patient/fluid column from electrical excitation circuitry is assured, even in the event of failure of the standard internal isolation.

This transducer is 56 mm (2.21 in.) long, with a maximum diameter at the base of 18 mm (0.71 in.). Its rated excitation voltage is 7.5 V which may be dc or an ac carrier.

Another type of transducer of smaller design is the P 50, shown in Figure 6.22. This unit can be mounted or attached to the patient near the measurement site (e.g., on the forearm of the patient). To achieve this miniaturization, the design has to be quite different. The sensing element of the P 50 is a tiny silicon beam upon which strain elements are diffused. This is, therefore, a bonded strain gage instead of the unbonded type used in the P 23 (see Section 2.3.1).

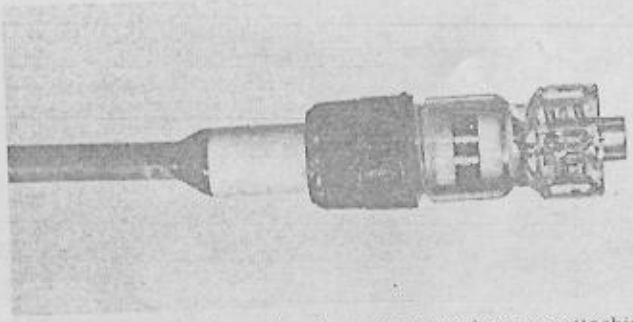


Figure 6.22. Bonded type blood pressure transducer for attaching to patient. (Courtesy of Gould Inc., Measurement Systems Division, Oxnard, CA.)

Both types of transducer described have pressure ranges of −50 to +300 mm Hg, with sensitivity for 7.5-V excitation of 50 μV/V/cm Hg.

These transducers can be flushed to remove air bubbles and to prevent blood from clotting at the end of the catheter. The fluid column used with a

transducer of this type has a natural frequency or resonance of its own that can affect the frequency response of the system. Care must be taken in selecting a transducer and catheter to be used together so that the frequency response of the complete system will be adequate. There are general-purpose models for arterial pressure (0 to 330 mm Hg) and venous pressure (0 to 50 mm Hg), and special models with differing sensitivities, volume displacement characteristics, and mechanical arrangements.

Pressure transducers are normally mounted on a suitable manifold near the patient's bed. It is important to keep the transducer at the same height as the point at which the measurements are to be made in order to avoid errors due to hydrostatic pressure differences. If a *differential pressure* is desired, two transducers of this type may be used at two different points, and the difference in pressure may be obtained as the difference of their output signals. Figure 6.23 shows a typical infusion manifold incorporating a transducer, a flushing system, and syringes for blood specimen withdrawal.

The signal-conditioning and display devices for these transducers are available in a variety of forms. However, each must provide a method of excitation for the strain-gage bridge, a means of zeroing or balancing the bridge, necessary amplification of the output signal, and a display device, such as a monitor scope, a recorder, panel meter, or digital readout device. Most modern systems permit many possible combinations.

Another type of blood pressure transducer is the *linear variable differential transformer* (LVDT) device, shown in an exploded view in Figure 6.24. Superficially, these transducers look similar to the unbonded strain-gage type. Indeed, with respect to the plastic dome used for visibility, the two pressure fittings for attachment to the catheter and for flushing, and the cable coming out of the bottom, they are similar. Such transducers also come in a variety of models with a range of characteristics for venous or arterial pressure, for different sensitivities, and for alternative volume displacements. The various models also have different natural frequencies and frequency responses.

It should be noted from the exploded view that these units disassemble into three subassemblies—the dome and pressure fittings subassembly, the center portion consisting of a stainless-steel diaphragm and core assembly, and the LVDT subassembly. There are two basic diaphragm and core assemblies with appropriate domes that are interchangeable in the coil-connector assembly. The first is used for venous and general-purpose clinical measurements and has a standard-size diaphragm with an internal fluid volume between the dome and diaphragm of less than 0.5 cm^3. The second design, with higher frequency response characteristics for arterial pressure contours, has a reduced diaphragm area and an internal volume of approximately 0.1 cm^3.

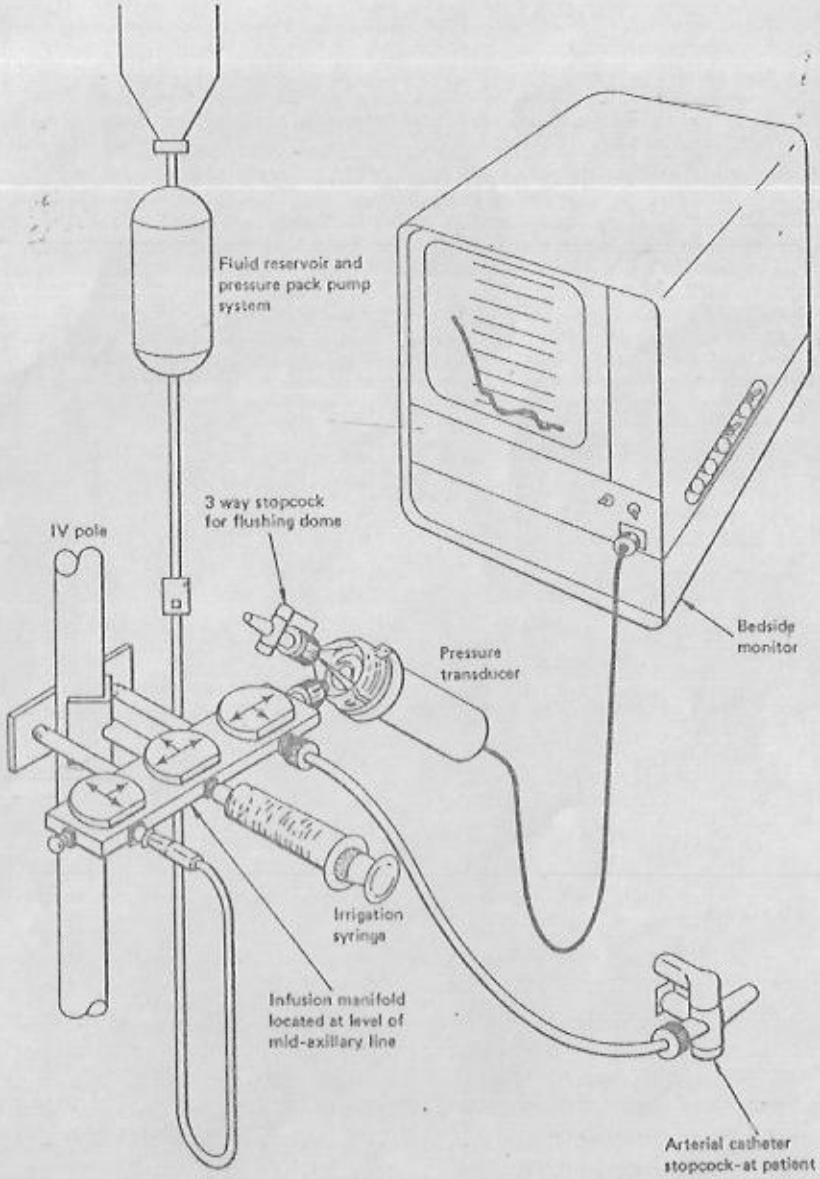

Fluid reservoir and pressure pack pump system
3 way stopcock for flushing dome
IV pole
Pressure transducer
Bedside monitor
Irrigation syringe
Infusion manifold located at level of mid-axillary line
Arterial catheter stopcock-at patient

Figure 6.23. Infusion manifold with transducer, flushing system and syringe. (Courtesy of Michael Tomeo, UCLA Medical Center.)

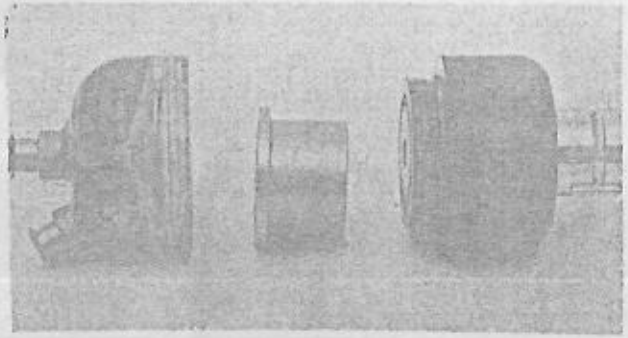


Figure 6.24. LVDT blood pressure transducer—exploded view. (Courtesy of Biotronex Laboratory, Inc., Silver Springs, MD.)

The Biotronex BL-9630 transducer is a linear variable differential transformer in which the primary coil is excited by an ac carrier (5 to 20 V peak to peak) in the range of 1500 to 15,000 Hz. Axial displacement of a movable iron core, attached to the diaphragm, cuts the magnetic lines of flux generated by the primary coil. Voltages induced in the secondary sensing coils are returned to the carrier amplifier, where they are differentially amplified and demodulated to remove the carrier frequency. The output of the carrier amplifier is a dc voltage proportional to diaphragm displacement. Linearity of the gage is better than ±1 percent of full range. Ordinary jarring and handling will not harm the gage. A positive mechanical stop is provided to prevent damage by as much as a 100 per-cent overpressure. The LVDT offers much higher signal levels than do conventional strain-gage transducers for a given excitation voltage.

6.2.4.2. Measurement at the site. To avoid the problems inherent in measuring blood pressure through a liquid column, a "catheter-tip" manometer can be fed through the catheter to the site at which the blood pressure is to be measured. This process requires a small-diameter transducer that is fairly rigid but flexible.

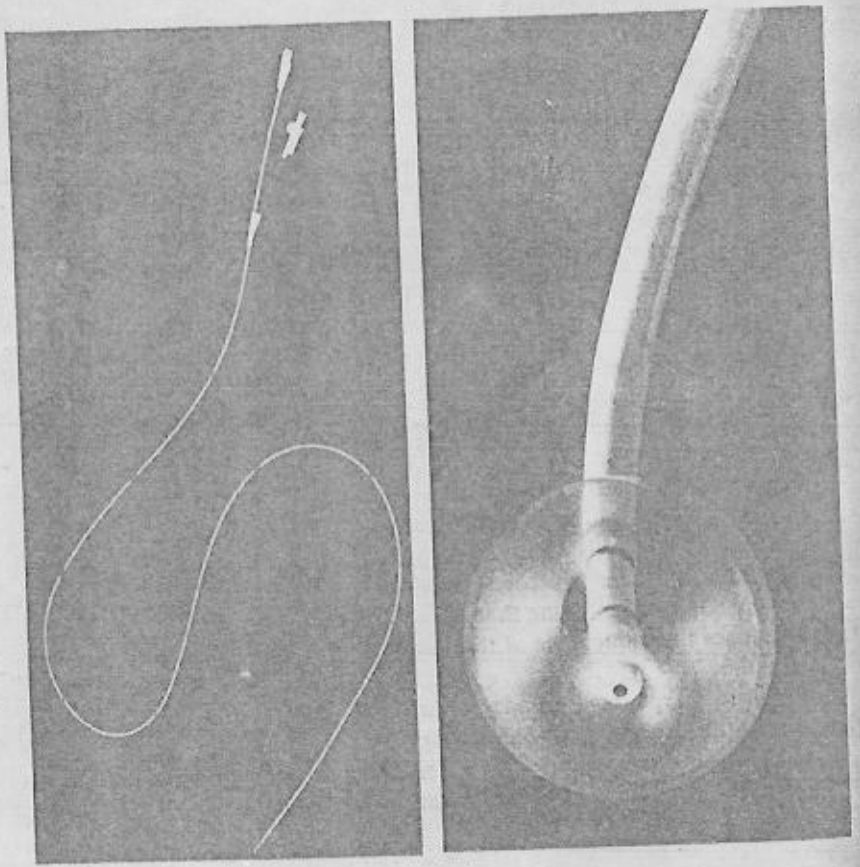
One such transducer makes use of the variable-inductance effect mentioned earlier. The tip is placed directly in the bloodstream so that the blood presses on a membrane surrounded by a protective cap. The membrane is connected to a magnetic slug that is free to move within a coil assembly and thus changes the inductance of the coil as a function of the pressure on the membrane.

Another type has a bonded strain-gage sensor built into the tip of a cardiac catheter. The resistance changes in the strain gage are a result of pressure variations at the site itself rather than through a fluid column. This gage can also be calibrated with a liquid-system catheter at the same location.

6.2.4.3. Floatation catheter. The construction of specialized, multiple-lumen "floatation" catheters has made insertion and continuous monitoring of pulmonary artery pressures feasible in most clinical settings. This type of catheter was designed by Drs. Swan and Ganz of the Cedars-Sinai Medical Center in Los Angeles and bears their names. Although specialized models are available, the basic catheter is approximately 110 cm in length and consists of a double-lumen tube with an inflatable balloon tip (Figure 6.25).

The catheter may be inserted percutaneously or via a venous cutdown. By using continuous pressure and electrocardiographic (ECG) monitoring,

Figure 6.25. (a) Swan-Ganz monitoring catheter for measurement of pulmonary artery and pulmonary capillary wedge pressures, full view; (b) balloon. (Courtesy of Edwards Laboratories, Division of American Hospital Supply Corporation, Santa Ana, CA.)



the catheter is threaded into the subclavian vein with the balloon deflated. At this point, the balloon is partially inflated to half capacity (0.4 to 0.6 cm³ of CO_2 or air) and carried downstream to the right atrium by the flow of blood. The balloon is then fully inflated (0.8 cm³) and advanced again so that the blood flow propels it through the tricuspid valve into the right ventricle. From there it is carried through the pulmonary valve into the pulmonary artery, where the balloon wedges in a distal artery branch. The position of the catheter is verified by the pressure tracing, which shifts from a pulmonary artery pressure indicator to the "wedged" pressure waveform position. Under ideal circumstances it should take the physician no more than 1 minute to float the catheter from full balloon inflation in the right atrium to the wedge position. During insertion, the fully inflated balloon covers the hard tip of the catheter, distributing pressure forces evenly across a broad area of the endocardium.

6.2.4.4. Percutaneous transducers. An example of a percutaneous blood pressure transducer is shown in Figure 6.26. It shows a transducer connected to a hypodermic needle that has been placed in a vessel of the arm. The three-way stopcock dome permits flushing of the needle, administering of drugs, and withdrawing of blood samples. This transducer can measure arterial or venous pressures, or the pressures of other physiological fluids, by direct attachment to a needle at the point of measurement. It can be used with a continuously self-flushing system without degradation of signal. The transparent plastic dome permits observation of air-bubble formation and consequent ejection. It is designed for use with a portable blood pressure monitor, which provides bridge excitation, balancing, and amplification. The meter scale is calibrated directly into millimeters of mercury. This transducer also has the advantage that it can be connected to a standard intravenous infusion bottle.

6.2.4.5. Implantable transducers. Figure 6.27 shows a type of transducer that can be implanted into the wall of a blood vessel or into the wall of the heart itself. This transducer is particularly useful for long-term investigations in animals.

The transducer's body is made of titanium, which has excellent corrosion-resistance characteristics, a relatively low thermal coefficient of expansion, and a low modulus of elasticity, which results in greater strain per unit stress. Four semiconductor strain gages are bonded to the inner surface of the pressure-sensing diaphragm. Transducers of this type come in a number of sizes (from 3 to 7 mm in diameter) for blood pressure measurement. A popular size is 4.5 mm in diameter. Larger sizes are available for pleural pressure. The thickness of the body is 1.2 to 1.3 mm in the various models.

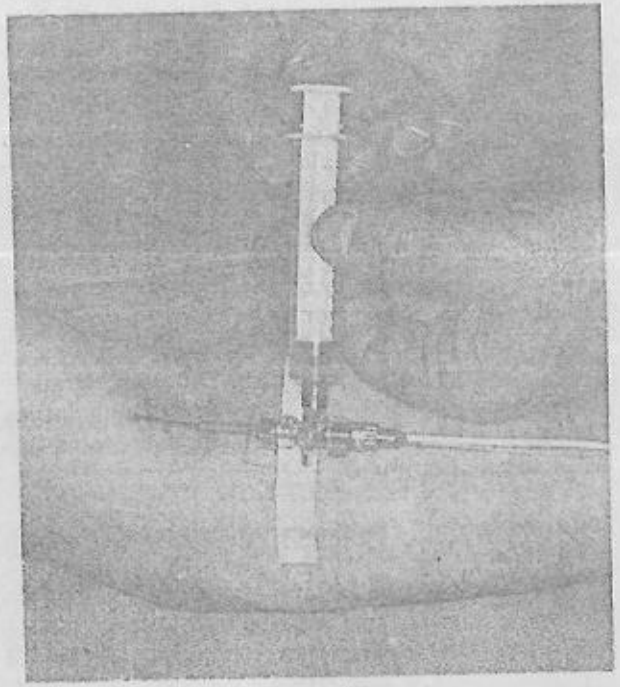
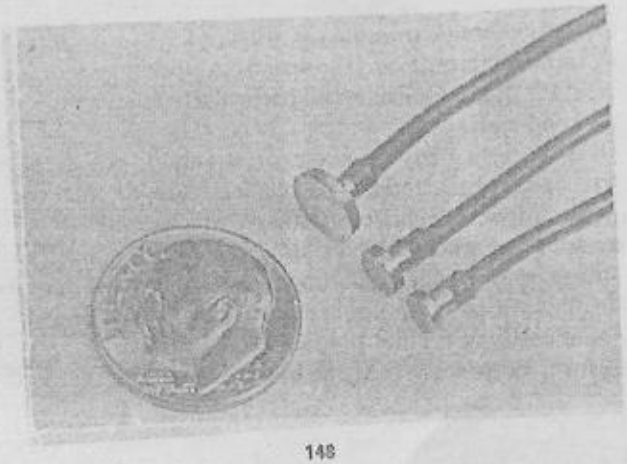


Figure 6.26. Percutaneous blood pressure measurement. Transducer in arm with three-way stopclock dome for administering drugs and withdrawing blood samples. (Courtesy of Gould Inc., Measurement Systems Division; Oxnard, CA.)

Figure 6.27. Implantable pressure transducer. (Courtesy of Konigsberg Instruments, Pasadena, CA.)



The four semiconductors are connected in bridge fashion as shown in Figure 6.19. As blood pressure increases on the diaphragm, the inner surface is stressed. The strain gages are located so that two of them are strained in tension while two are in compression. When the bridge is excited, an output voltage proportional to the blood pressure can be obtained.

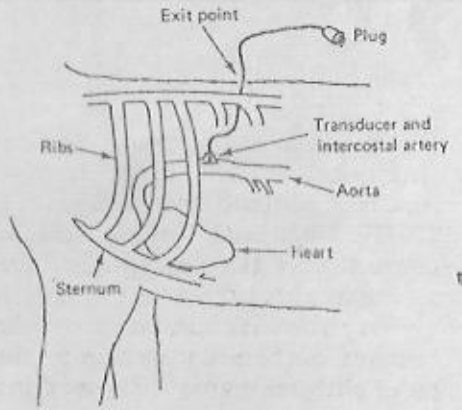
Additional resistors, connected externally to the bridge, provide temperature compensation, although these bridges are not extremely sensitive to temperature. Since they operate in the bloodstream at a fairly constant 37 °C, the temperature effects are not serious.

These transducers can be excited with ac or dc and easily lend themselves to telemetry application. In service, they have proven very reliable. Cases of chronic implants (in excess of 2 years) have been reported with no detrimental effect on the animal, the gage, or the wires. The wires are usually insulated with a plastic compound, polyvinylchloride, which is fairly impervious to body fluids.

There are many examples of the use of this type of transducer in animal research, including the implantation in both ventricles of the heart, the aorta, the carotid artery, and the femoral artery. In addition to blood pressures, they have also been used for measuring abdominal, esophageal, thoracic, intrauterine and intracranial pressures.

To implant a transducer in an artery, a longitudinal incision is made; the transducer is inserted with its housing in intimate contact with the arterial walls. The wound is closed with interrupted sutures. For cardiac implants, a stab wound in the ventricle permits ready insertion, with the transducer placed free of both the myocardium and (in the left ventricle) the chordae tendonae. A technique used for long-term studies of the blood pressure in the aorta is to insert the transducer from the opposite side and use a small intercostal artery to bring the wire through. This creates a

Figure 6.28. Transducer implanted in the aorta.



stronger bind in active animals. The wire is held to the artery by a purse-string suture. Figure 6.28 shows such a preparation. In this case the plug was inserted into a biotelemetry transmitter so that the blood pressure data could be received remotely. The use of these transducers and telemetry have been useful in gathering information on exercise, the effect of drugs and extreme environments, and acceleration and impact studies (see Chapter 12).

6.3. MEASUREMENT OF BLOOD FLOW AND CARDIAC OUTPUT

An adequate blood supply is necessary for all organs of the body; in fact, an impaired supply of blood is the cause of various diseases. The ability to measure blood flow in the vessel that supplies a particular organ would therefore be of great help in diagnosing such diseases. Unfortunately, blood flow is a rather elusive variable that cannot be measured easily.

The rate of flow of a liquid or gas in a pipe is expressed as the volume of the substance that passes through the pipe in a given unit of time. Flow rates are therefore usually expressed in liters per minute or milliliters per minute (cm³/min).

Methods used in industry for flow measurements of other liquids, like the turbine flowmeter and the rotameter, are not very suitable for the measurement of blood flow because they require cutting the blood vessel. These methods also expose the blood to sharp edges, which are conducive to blood-clot formation.

Practically all blood flow meters currently used in clinical and research applications are based on one of the following physical principles:

1. Electromagnetic induction.
2. Ultrasound transmission or reflection.
3. Thermal convection.
4. Radiographic principles.
5. Indicator (dye or thermal) dilution.

Magnetic and *ultrasonic blood flow meters* actually measure the *velocity* of the bloodstream. Because these techniques require that a transducer surround an excised blood vessel, they are mainly used during surgery. Ultrasound, however, can be used transcutaneously to detect obstructions of blood vessels where quantitative blood flow measurements are not required.

A *plethysmograph,* which actually indicates volume changes in body segments, can be used to measure the flow of blood in the limbs. The principle of plethysmography is discussed in Section 6.4.

6.3.1. Magnetic Blood Flow Meters.

Magnetic blood flow meters are based on the principle of magnetic induction. When an electrical conductor is moved through a magnetic field, a voltage is induced in the conductor proportional to the velocity of its motion. The same principle applies when the moving conductor is not a wire, but rather a column of conductive fluid that flows through a tube located in the magnetic field. Figure 6.29 shows how this principle is used in magnetic blood flow meters. A permanent magnet or electromagnet positioned around the blood vessel generates a magnetic field perpendicular to the direction of the blood flow. The voltage induced in the moving blood column is measured with stationary electrodes located on opposite sides of the blood vessel and perpendicular to the direction of the magnetic field.

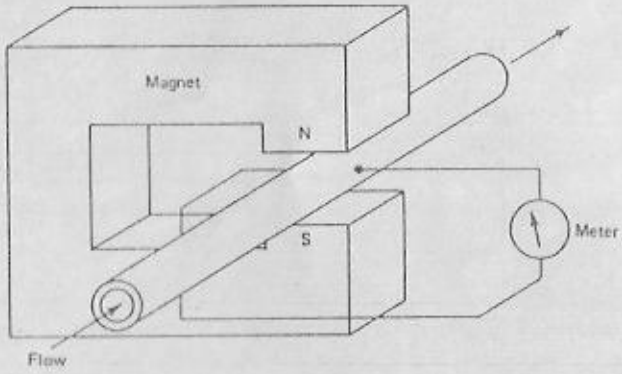


Figure 6.29. Magnetic blood flow meter, principle.

The most commonly used types of implantable magnetic blood flow probes are shown in Figures 6.30 through 6.32. The *slip-on* or *C type* is applied by squeezing an excised blood vessel together and slipping it through the slot of the probe. In some transducer models the slot is then closed by inserting a keystone-shaped segment of plastic, as shown. Contact is provided by two slightly protruding platinum disks that touch the wall of the blood vessel. For proper operation, the orifice of the probe must fit tightly around the vessel. For this reason, probes of this type are manufactured in sets, with diameters increasing in steps of 0.5 or 1 mm from about 2 to 20 mm. The probes shown in Figure 6.30 can be implanted for chronic use. In contrast, Figure 6.31 shows a model with a long handle for use during surgery.

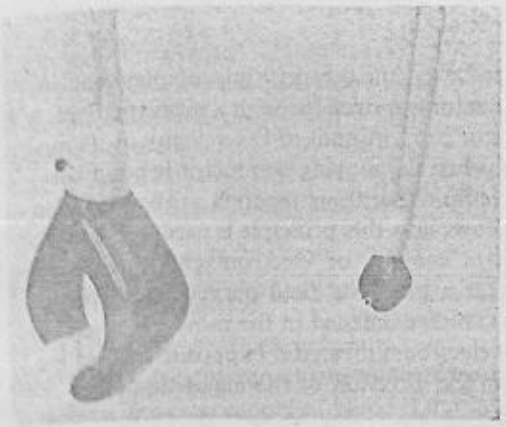


Figure 6.30. Samples of large and small lumen diameter blood flow probes. (Courtesy of Micron Instruments, Los Angeles, CA.)

Figure 6.31. Blood flow probe—clip-on type for use during surgery. (Courtesy of Biotronex, Silver Springs, MD.)

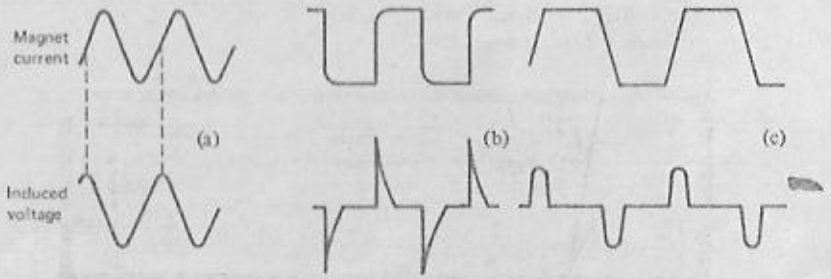
In the *cannula-type transducer,* the blood flows through a plastic cannula around which the magnet is arranged. The contacts penetrate the walls of the cannula. This type of transducer requires that the blood vessel be cut and its ends slipped over the cannula and secured with a suture. A similar type of transducer (Figure 6.32) is also used to measure the blood flow in extracorporeal devices, such as dialyzers. Magnetic blood flow meters actually measure the mean blood velocity. Because the cross-sectional area at the place of velocity measurement is well defined with either type of transducer, these transducers can be calibrated directly in units of flow.

Figure 6.32. Extracorporeal blood flow probe. (Courtesy of Biotronex, Silver Springs, MD.)

Magnetic blood flow transducers are also manufactured as catheter-tip transducers. For this type, the normal transducer design is essentially turned "inside out," with the electromagnet being located inside the catheter, which has the electrodes at the outside. Catheter transducers cannot be calibrated in flow units, however, because the cross section of the blood vessel at the place of measurement is not defined.

The output voltage of a magnetic blood flow transducer is very small, typically in the order of a few microvolts. In early blood flow meters, a constant magnetic field was used, which caused difficulties with electrode polarization and amplifier drift. To overcome these problems, all contemporary magnetic blood flow meters use electromagnets that are driven by alternating currents. Doing this, however, creates another problem: the change of the magnetic field causes the transducer to act like a transformer and induces error voltages that often exceed the signal levels by several orders of magnitude. Thus, for recovering the signal in the presence of the error voltage, amplifiers with large dynamic range and phase-sensitive or gated detectors have to be used. To minimize the problem, several different waveforms have been advocated for the magnet current, as shown in Figure 6.33. With a sinusoidal magnet current, the induced voltage is also

Figure 6.33. Waveforms used in magnectic blood flow meters and error signals induced by the current: (a) sine wave; (b) square wave; (c) trapezoidal wave.



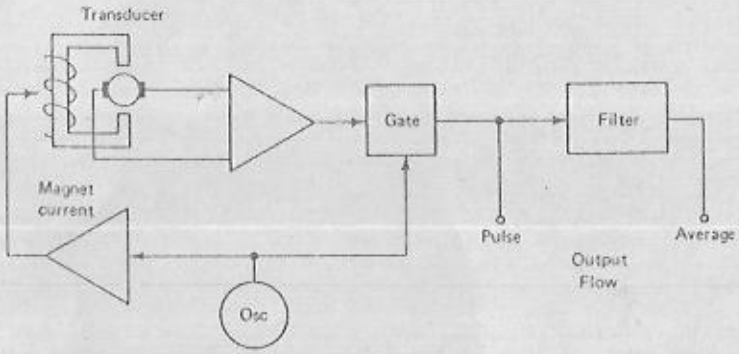


Figure 6.34. Magnetic blood flow meter, block diagram.

sinusoidal but is 90° out of phase with the flow signal. With a suitable circuit, similar to a bridge, the induced voltage can be partially balanced out. With the magnet current in the form of a square wave, the induced voltage should be zero once the spikes from the polarity reversal have passed. In practice, however, these spikes are often of extremely high amplitude, and the circuitry response tends to extend their effect. A compromise is the use of a magnet current having a trapezoidal waveform. None of the three waveforms used seems to have demonstrated a definite superiority.

The block diagram of a magnetic blood flow meter is shown in Figure 6.34. The oscillator, which drives the magnet and provides a control signal for the gate, operates at a frequency of between 60 and 400 Hz. The use of a gated detector makes the polarity of the output signal reverse when the flow direction reverses. The frequency response of this type of system is usually high enough to allow the recording of the flow pulses, while the mean or average flow can be derived by use of a low-pass filter. Figure 6.35 shows a single-channel magnetic blood flow meter that can be used with a variety of different transducers.

Figure 6.35. Magnetic blood flow meter. (Courtesy of Micron Instruments, Los Angeles, CA.)

6.3.2. Ultrasonic Blood Flow Meters.

In an *ultrasonic blood flow meter,* a beam of ultrasonic energy is used to measure the velocity of flowing blood. This can be done in two different ways. In the *transit time ultrasonic flow meter,* a pulsed beam is directed through a blood vessel at a shallow angle and its transit time is then measured. When the blood flows in the direction of the energy transmission, the transit time is shortened. If it flows in the opposite direction, the transit time is lengthened.

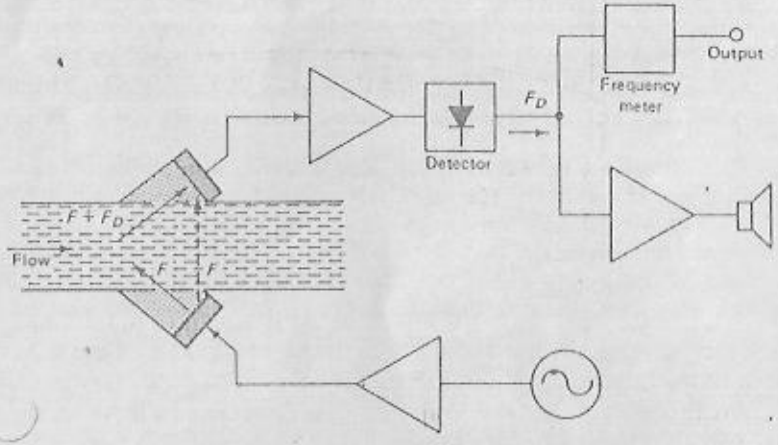


Figure 6.36. Ultrasonic blood flow meter, Doppler type.

More common are ultrasonic flow meters based on the Doppler principle (Figure 6.36). An oscillator, operating at a frequency of several megahertz, excites a piezoelectric transducer (usually made of barium titanate). This transducer is coupled to the wall of an exposed blood vessel and sends an ultrasonic beam with a frequency F into the flowing blood. A small part of the transmitted energy is scattered back and is received by a second transducer arranged opposite the first one. Because the scattering occurs mainly as a result of the moving blood cells, the reflected signal has a different frequency due to the Doppler effect. Its frequency is either $F + F_D$ or $F - F_D$, depending on the direction of the flow. The Doppler component F_D is directly proportional to the velocity of the flowing blood. A fraction of the transmitted ultrasonic energy, however, reaches the second transducer directly, with the frequency being unchanged. After amplification of the composite signal, the Doppler frequency can be obtained at the output of a detector as the difference between the direct and the scattered signal components.

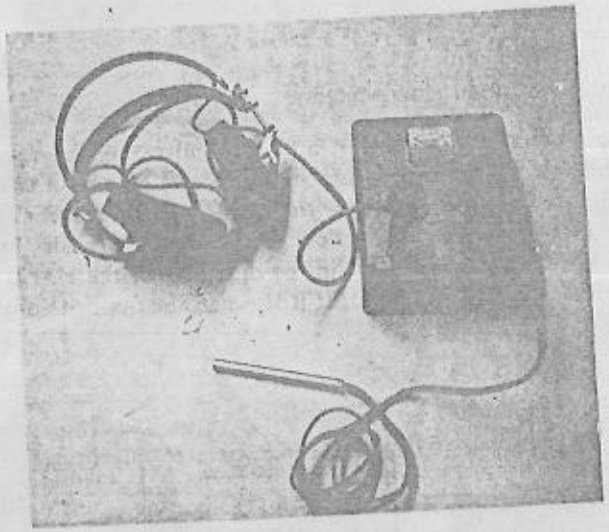


Figure 6.37. Percutaneous Doppler device with probe and earphones. (Courtesy of Veterans Administration Biomedical Engineering and Computing Center, Sepulveda, CA.)

With blood velocities in the range normally encountered, the Doppler signal is typically in the low audio frequency range. Because of the velocity profile of the flowing blood, the Doppler signal is not a pure sine wave, but has more the form of narrow-band noise. Therefore, from a loudspeaker or earphone, the Doppler signal of the pulsating blood flow can be heard as a characteristic "swish—swish—." When the transducers are placed in a suitable mount (which defines the area of the blood vessel), a frequency meter used to measure the Doppler frequency can be calibrated directly in flow-rate units. Unfortunately, Doppler flow meters of this simple design cannot discriminate the direction of flow. More complicated circuits, however, which use the insertion of two quadrature components of the carrier, are capable of indicating the direction of flow.

Transducers for ultrasonic flow meters can be implanted for chronic use. Some commercially available flow meters of this type incorporate a telemetry system to measure the blood flow in unrestrained animals.

Figure 6.37 is an illustration of a simple Doppler device, with the two transducers mounted in a hand-held probe which is now widely used to trace blood vessels close to the surface and to determine the location of vascular obstructions. In order to facilitate transmission of ultrasonic energy, the probe must be coupled to the skin with an aqueous jelly. Such devices can be used to detect the motion of internal structures in the body—for example, the fetal heart. (See Chapter 9 for further discussion of ultrasonic diagnosis.)

6.3.3. Blood Flow Measurement by Thermal Convection

A hot object in a colder-flowing medium is cooled by thermal convection. The rate of cooling is proportional to the rate of the flow of the medium. This principle, often used to measure gas flow, has also been applied to the measurement of blood velocity. In one application, a thermistor in the bloodstream is kept at a constant temperature by a servo system. The electrical energy required to maintain this constant temperature is a measure of the flow rate. In another method an electric heater is placed between two thermocouples or thermistors that are located some distance apart along the axis of the vessel. The temperature difference between the upstream and the downstream sensor is a measure of the blood velocity. A device of the latter type is sometimes called *a thermostromuhr* (literally, from the German "heat current clock"). Thermal convection methods for blood flow determination, although among the oldest ones used for this purpose, have now been widely replaced by the other methods described in this chapter.

6.3.4. Blood Flow Determination by Radiographic Methods

Blood is not normally visible on an X-ray image because it has about the same radio density as the surrounding tissue. By the injection of a contrast medium into a blood vessel (e.g., an iodated organic compound), the circulation pattern can be made locally visible. On a sequential record of the X-ray image (either photographic or on a videotape recording), the progress of the contrast medium can be followed, obstructions can be detected, and the blood flow in certain blood vessels can be estimated. This technique, known as *cine* (or *video*) *angiography*, can be used to assess the extent of damage after a stroke or heart attack.

Another method is the injection of a radioactive isotope into the blood circulation, which allows the detection of vascular obstructions (e.g., in the lung) with an imaging device for nuclear radiation, such as a scanner or gamma camera (see Chapter 14).

Vascular obstructions in the lower extremities can sometimes be detected by measuring differences in the skin temperature caused by the reduced circulation. This can be accomplished by one of the various methods of skin surface temperature measurement described in Chapter 9.

6.3.5. Measurement by Indicator Dilution Methods

The indicator or dye dilution methods are the only methods of blood flow measurement that really measure the blood flow and not the blood velocity. In principle, any substance can be used as an indicator if it mixes readily with blood and its concentration in the blood can be easily determined after

mixing. The substance must be stable but should not be retained by the body. It must have no toxic side effects.

An indocyanine dye, Cardiogreen, used in an isotonic solution was long favored as a indicator. Its concentration was detemined by measuring the light absorption with a densitometer (colorimeter). Radioactive isotopes (radioiodited serum albumen) have also been employed for this purpose. The indicator most frequently used today, however, is isotonic saline, which is injected at a temperature lower than the body temperature. The concentration of the saline after mixing with the blood is determined with a sensitive thermistor thermometer (see Chapter 9).

The principle of the dilution method is shown in Figure 6.38. The upper left drawing shows a model of a part of the blood circulation under the (very simplified) assumption that the blood is not recirculated. The indicator is injected into the flow continuously, beginning at time t, at a constant infusion rate I (grams per minute). A detector measures the concentration downstream from the injection point. Figure 6.38 (a) shows the output of a recorder that is connected to the detector. At a certain time after the injection, the indicator begins to appear, the concentration increases, and, finally, it reaches a constant value, C_0 (milligrams per liter). From the measured concentration and the known injection rate, I (in milligrams per minute), the flow can be calculated as

$$F \text{ (liters per minute)} = \frac{I \text{ (milligrams per minute)}}{C_0 \text{ (milligrams per liter)}}$$

The earliest method for determining cardiac output, the *Fick method,* is based on this simple model. The indicator is the oxygen of the inhaled air that is "injected" into the blood in the lungs. The "infusion rate" is determined by measuring the oxygen content of the exhaled air and subtracting it from the known oxygen content of the inhaled room air. The oxygen metabolism only approximately resembles the model of the open circulation, because only part of the oxygen is consumed in the systemic circulation and the returning venous blood still contains some oxygen. Therefore, the oxygen concentration in the returning venous blood has to be determined and subtracted from the oxygen concentration in the arterial blood leaving the lungs. The measurements are averaged over several minutes to reduce the influence of short-term fluctuations. An automated system is available that measures the oxygen concentration (by colorimetry) and the oxygen consumption, and continuously calculates the cardiac output from these measurements.

When a dye or isotope is used as an indicator, the concentration does not assume a steady-state value but increases in steps whenever the recirculated indicator again passes the detector [points R in Figure 6.38 (b)].

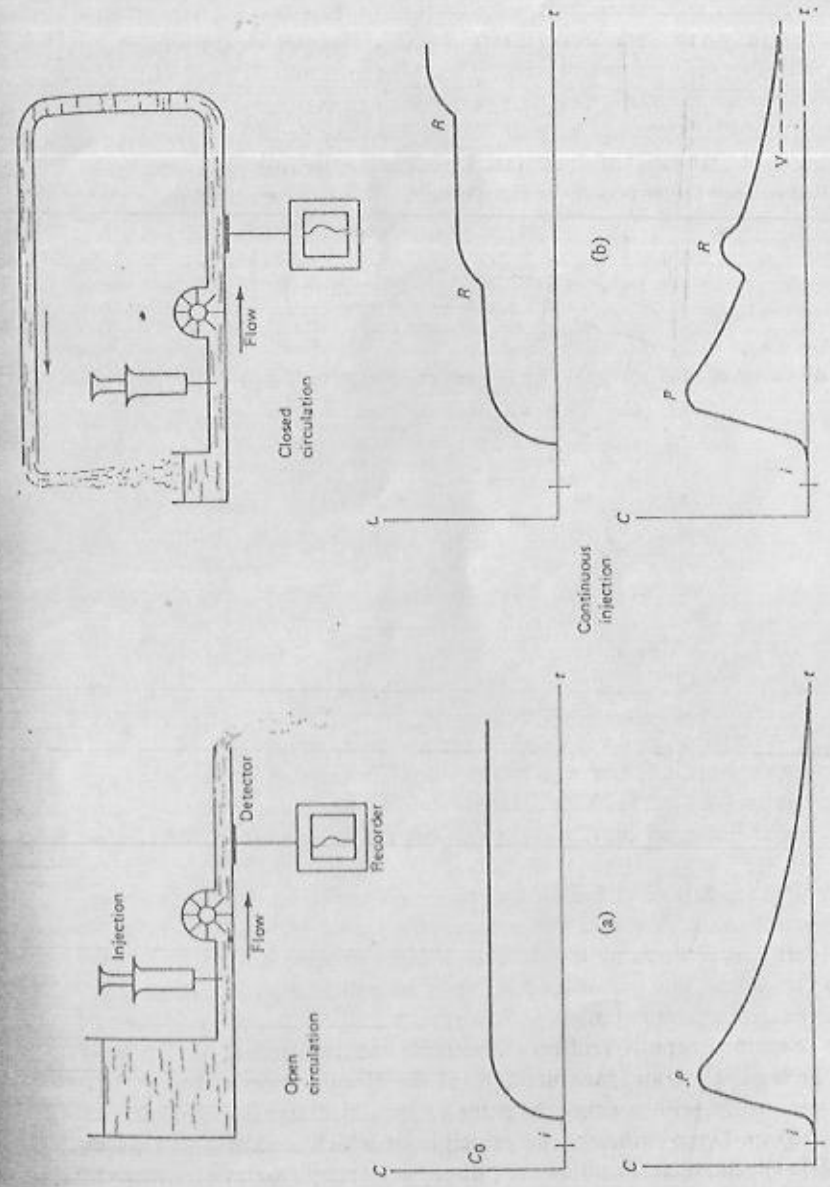


Figure 6.38. Flow measurements by indicator dilution methods, principle. (The indicator is injected at time $t = i$.)

The recirculation often occurs before the concentration has reached a plateau. Consequently, a slightly different method is usually used, and the indicator is injected as a bolus instead of being infused at a constant rate. Figure 6.38(c) shows the concentration for this case, again under the assumption that it is an open system. The concentration increases at first, reaches a peak, P, and then decays as an exponential function. This "washout curve" is mainly a result of the velocity profile of the blood, which causes a "spread" of the bolus. To calculate the flow, the area under the concentration curve has to be determined. This is given by the integral:

$$\int_{t=i}^{t=\infty} C\,dt \quad \left(\frac{\text{milligrams}}{\text{liter}} \times \text{minutes}\right)$$

From the value of this integral, and from the amount B of the injected indicator (in milligrams), the flow can be calculated:

$$F \left(\frac{\text{liters}}{\text{minute}}\right) = \frac{B}{\int_{t=i}^{t=\infty} C\,dt} \left(\frac{\text{milligrams}}{\text{milligrams/liter} \times \text{minutes}}\right)$$

Because of the recirculation, the concentration does not show a monotonous decay as in Figure 6.38(b); instead, after some time, a "hump" (R in the figure) occurs. Normally, the portion of the curve preceding the recirculation hump is exponential, and the decay curve can be determined, despite the recirculation, by exponential extrapolation (*Hamilton method*). This was originally done by manually replotting the curve on semilogarithmic paper, which resulted in a straight line for the exponential part of the curve. This line was then extended, the extended part replotted on the original plot, and the extrapolated curve integrated with a planimeter.

To simplify this complicated and time-consuming operation, various special-purpose analog computers were used which employed several different algorithms or determining the area under the curve despite the recirculation distortion. The use of cold saline as an indicator avoids these problems. Because the volume injected is normally only 10 ml, the circulating blood is rewarmed rapidly and no measurable recirculation occurs. Injection of the cool saline and measurement of the blood temperature are frequently performed with a single catheter of special design. This device is called the *Swan-Ganz catheter,* the principle of which is shown in Figure 6.39. It is a special adaptation of the pulmonary artery floatation catheter described in Section 6.2.4.3. This catheter contains four separate lumens. One lumen terminates about 30 cm (12 in.) from the tip and is used to inject

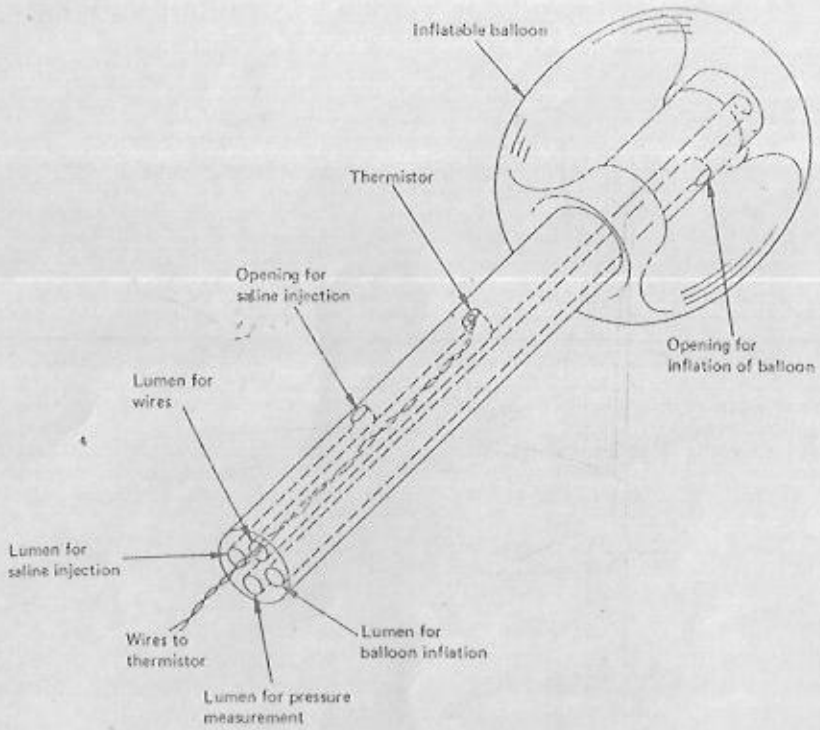
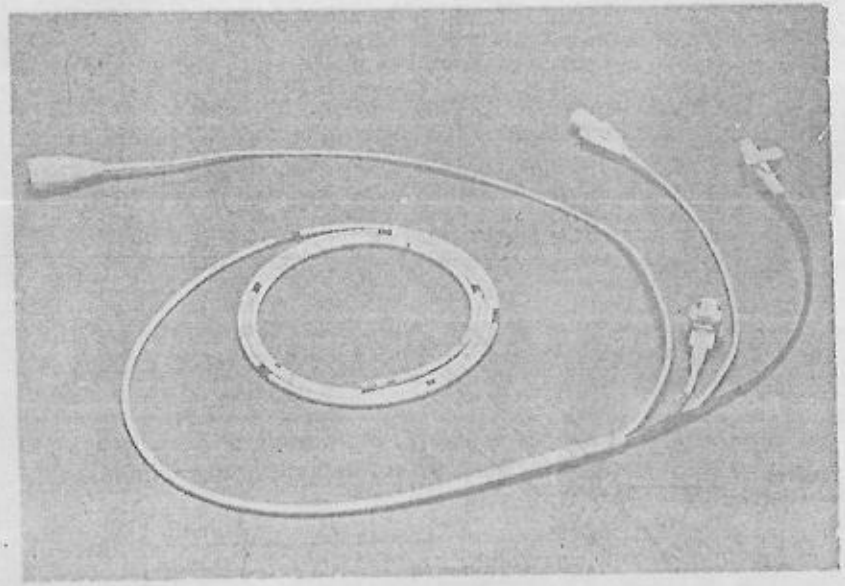


Figure 6.39. Swan-Ganz catheter for the measurement of cardiac output by the thermal dilution method. (Not drawn to scale.) The opening for saline injection is actually located 30 cm distal from the catheter tip and the lumens vary in diameter.

the cooled saline. The second lumen contains two thin wires that lead to a tiny electrical temperature sensor close to the top of the catheter. The third lumen ends at the catheter tip and can be used to measure the blood pressure at this point with one of the pressure transducers described in Section 6.2. The fourth lumen is used to inflate a small rubber ballon at the tip of the catheter. Once the catheter has been inserted into a vein, the ballon is inflated and the returning venous blood carries the catheter until its tip is positioned in the pulmonary artery. The position of the catheter can be checked by measuring the pressure at its tip. Thus, the catheter can, if necessary, be inserted without fluoroscopic control.

The thermistor is connected into a bridge circuit which permits measurement and recording of the blood temperature during injection. A relatively simple analog computer, which consists essentially of an electronic integrator and necessary controls, permits direct reading of the cardiac output. The complete thermodilution catheter and the cardiac output computer are illustrated in Figure 6.40.



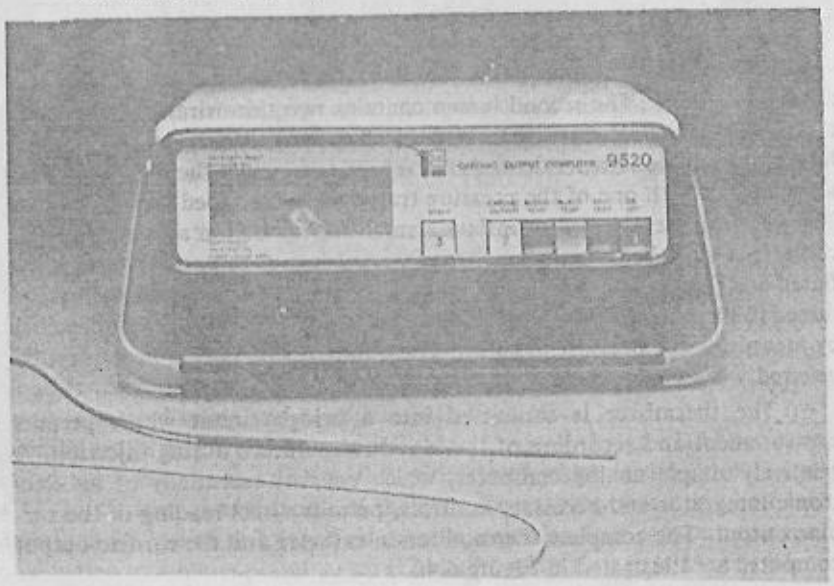


Figure 6.40. Cardiac output catheter and computer: (a) complete thermodilution Swan-Ganz floatation catheter; (b) cardiac output computer. (Courtesy of Edwards Laboratories, Division of American Hospital Supply Corporation, Santa Ana, CA.)

6.4. PLETHYSMOGRAPHY

Related to the measurement of blood flow is the measurement of volume changes in any part of the body that result from the pulsations of blood occuring with each heartbeat. Such measurements are useful in the diagnosis of arterial obstructions as well as for pulse-wave velocity measurements. Instruments measuring volume changes or providing outputs that can be related to them are called *plethysmographs,* and the measurement of these volume changes, or phenomena related thereto, is called *plethysmography*.

A "true" plethysmograph is one that actually responds to changes in volume. Such an instrument consists of a rigid cup or chamber placed over the limb or digit in which the volume changes are to be measured, as shown in Figure 6.41. The cup is tightly sealed to the member to be measured so that any changes of volume in the limb or digit reflect as pressure changes inside the chamber. Either fluid or air can be used to fill the chamber.

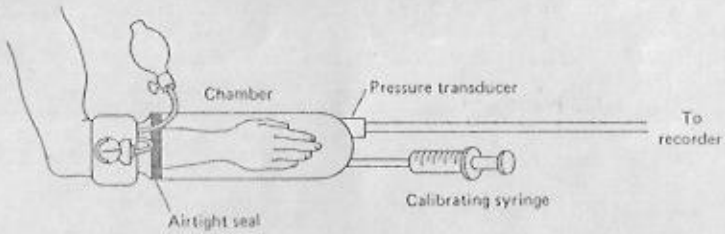


Figure 6.41. Plethysmograph. (Redrawn from A.C. Guyton, *Textbook of Medical Physiology,* 4th ed., W.B. Saunders Co., 1971, by permission.)

Plethysmographs may be designed for constant pressure or constant volume within the chamber. In either case, some form of pressure or displacement transducer must be included to respond to pressure changes within the chamber and to provide a signal that can be calibrated to represent the volume of the limb or digit. (See the description of the pressure transducers in Section 6.2.2. and displacement transducers in Chapter 2.) The baseline pressure can be calibrated by use of a calibrating syringe.

This type of plethysmograph can be used in two ways (see Figue 6.41). If the cuff, placed upstream from the seal, is not inflated, the output signal is simply a sequence of pulsations proportional to the individual volume changes with each heartbeat.

The plethysmograph illustrated in Figure 6.41 can also be used to measure the total amount of blood flowing into the limb or digit being measured. By inflating the cuff (placed slightly upstream from the seal) to a

pressure just above venous pressure, arterial blood can flow past the cuff, but venous blood cannot leave. The result is that the limb or digit increases its volume with each heartbeat by the volume of the blood entering during that beat. The output tracing for this measurement is shown in Figure 6.42. The slope of a line along the peaks of these pulsations represents the overall rate at which blood enters the limb or digit. Note, however, that after a few seconds the slope tends to level off. This is caused by a back pressure that builds up in the limb or digit from the accumulation of blood that cannot escape.

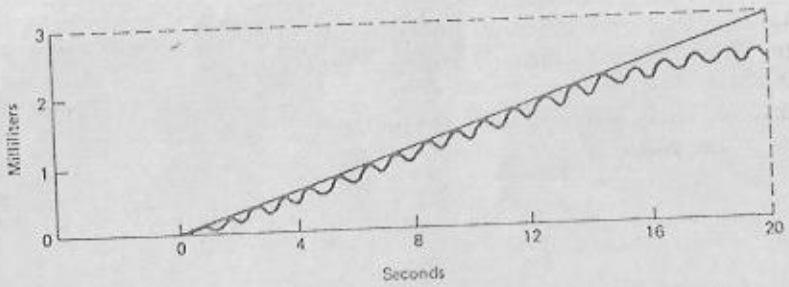


Figure 6.42. Blood volume record from plethysmograph. (From A.C. Guyton, *Textbook of Medical Physiology*, 4th ed., W.B. Saunders Co., 1971, by permission.)

Another device that quite closely approximates a "true" plethysmograph is the *capacitance plethysmograph* shown in Figure 6.43. In this device, which is generally used on either the arm or leg, the limb in which the volume is being measured becomes one plate of a capacitor. The other plate is formed by a fixed screen held at a small distance from the limb by an insulating layer. Often a second screen surrounds the outside plate at a fixed distance to act as a shield for greater electrical stability. Pulsations of the blood in the arm or leg cause variations in the capacitance, because the distance between the limb and the fixed screen varies with these pulsations. Some form of capacitance-measuring device is then used to obtain a continuous measure of these variations. Since the length of the cuff is fixed, the variations in capacitance can be calibrated as volume variations. The device can be calibrated by using a special cone of known volume on which the diameter can be adjusted to provide the same capacitance reading as the limb on which measurements are made. Since the capacitance plethysmograph essentially integrates the diameter changes over a segment of the limb, its readings are reasonably close to those of a "true" plethysmograph. Also, as with the "true" plethysmograph, estimates of the total volume of blood entering an arm or leg over a given period of time can be made by placing an occluding cuff just upstream from the capacitance device and by

Figure 6.43. Capacitance plethysmograph. (Designed by one of the authors for V.A. Hospital, San Francisco, CA.)

pressurizing the cuff to a pressure greater than venous pressure but below arterial pressure.

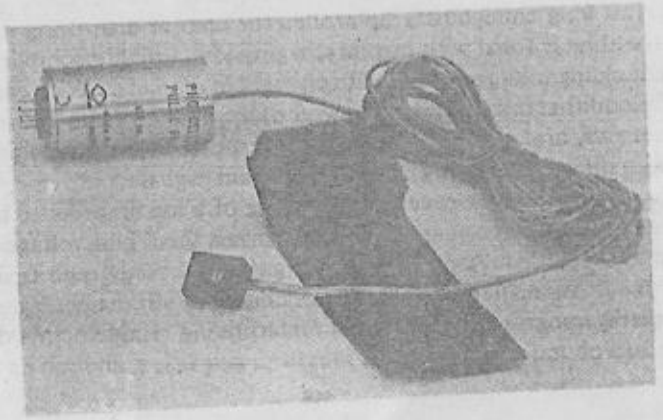
Several devices, called plethysmographs, actually measure some variable related to volume rather than volume itself. One class of these "pseudo-plethysmographs" measures changes in diameter at a certain cross section of a finger, toe, arm, leg, or other segment of the body. Since volume is related to diameter, this type of device is sufficiently accurate for many purposes.

A common method of sensing diameter changes is through the use of a *mercury strain gage,* which consists of a segment of small-diameter elastic tubing, just long enough to wrap around the limb or digit being measured. When the tube is filled with mercury, it provides a highly compliant strain gage that changes its resistance with changes in diameter. With each pulsation of blood that increases the diameter of the limb or digit, the strain gage elongates and, in stretching, becomes thinner, thus increasing its resistance. The major difficulty in using the mercury strain gage is its extremely low impedance. This drawback necessitates the use of a low-impedance bridge to measure small resistance variations and convert them into voltage changes that can be recorded. A mercury strain gage plethysmograph is shown in Figure 6.44. A difficulty that is common to all diameter-measuring pseudoplethysmographs is that of interpreting single-point diameter changes as volume changes.

Figure 6.44. Mercury strain gage plethysmograph. (Courtesy of Parks Electronics Laboratory, Beaverton, OR.)

Another step away from the true plethysmograph is the *photoelectric plethysmograph*. This device operates on the principle that volume changes in a limb or digit result in changes in the optical density through and just beneath the skin over a vascular region. A photoelectric plethysmograph is shown in Figure 6.45. A light source in an opaque chamber illuminates a small area of the fingertip or other region to which the transducer is applied. Light scattered and transmitted through the capillaries of the region is picked up by the photocell, which is shielded from all other light. As the capillaries fill with blood (with each pulse), the blood density increases, thereby reducing the amount of light reaching the photocell. The result causes resistance changes in the photocell that can be measured on a Wheatstone bridge and recorded. Pulsations recorded in this manner are

Figure 6.45. Photoelectric plethysmograph. (Courtesy of Narco BioSystems, Inc., Houston, TX.)



somewhat similar to those obtained by a true plethysmograph, but the photocell device cannot be calibrated to reflect absolute or even relative volumes. As a result, this type of measurement is primarily limited to detecting the fact that there are pulsations into the finger, indicating heart rate and determining the arrival time of the pulses. One serious difficulty experienced with this type of device is the fact that even the slightest movement of the finger with respect to the photocell or light source results in a severe amount of movement artifact. Furthermore, if the light source produces heat, the effect of the heat may change local circulation beneath the light source and photocell.

A more reliable device is the *impedance plethysomgraph,* in which volume changes in a segment of a limb or digit are reflected as impedance changes. These impedance changes are due primarily to changes in the conductivity of the current path with each pulsation of blood. Impedance plethysmographic measurements can be made using a two-electrode or a four-electrode system. The electrodes are either conductive bands wrapped around the limb or digit to be measured or simple conductive strips of tape attached to the skin. In either case, the electrodes contact the skin through a suitable electrolyte jelly or paste to form an electrode interface and to remove the effect of skin resistance. In a two-electrode system, a constant current is forced through the tissue between the two electrodes, and the resulting voltage changes are measured. In the four-electrode system, the constant current is forced through two outer, or current electrodes, and the voltage between the two inner, or measurement, electrodes is measured. The internal body resistances between the electrodes form a physiological voltage divider. The advantage of the four-electrode system is a much smaller amount of current through the measuring electrodes, thus reducing the possiblility of error due to changes in electrode resistance. Currents used for impedance plethysmography are commonly limited to the low-microampere range. The driving current is ac, sometimes a square wave, and usually of a high-enough frequency (around 10 kHz or higher) to reduce the effect of skin resistance. At these frequencies the capacitive component of the skin electrode interface becomes a significant factor.

Several theories attempt to explain the actual cause of the measured impedance changes. One is that the mere presence of additional blood filling a segment of the body lowers the impedance of that segment. Tests reported by critics of this method, however, claim that the actual impedance difference between the blood-filled state and more "empty" state is not significant.

A second theory is that the increase in diameter due to additional blood in a segment of the body increases the cross-sectional area of the segment's conductive path and thereby lowers the resistance of the path. This may be true to some extent, but again the percentage of area change is very small.

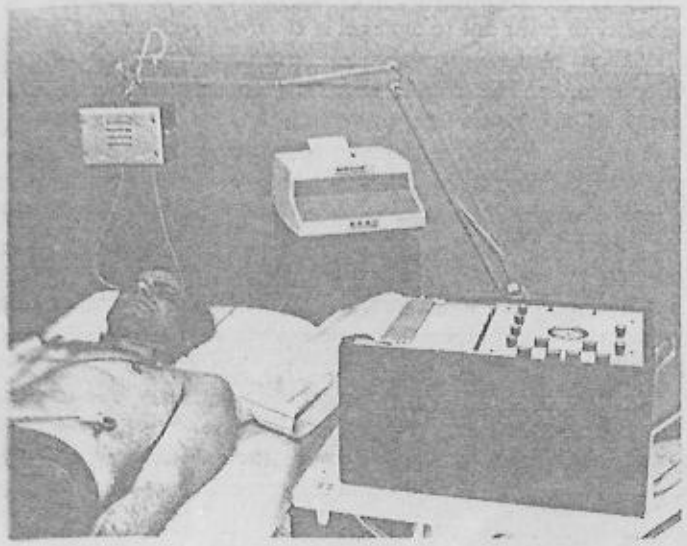
Critics of impedance plethysmography argue that the measured impedance changes are actually changes in the impedance of the skin-electrode interface, caused by pressure changes on the electrodes that occur with each blood pulsation.

Whatever the reason, however, impedance plethysmography does produce a measure that closely approximates the output of a true plethysmograph. Its main difficulty is the problem of relating the output resistance to any absolute volume measurement. As with the photocell plethysmograph, detection of the presence of arterial pulsations, measurment of pulse rate, and determination of time of arrival of a pulse at any given point in the peripheral circulation can all be satisfactorily handled by impedance plethysmography. Also, the impedance plethysmograph can measure time-variant changes in blood volume.

A special form of impedance plethysmography is *rheoencephalography*, the measurement of impedance changes between electrodes positioned on the scalp. Although primarily limited to research applications, this technique provides information related to cerebral blood flow and is sometimes used to detect circulatory differences between the two sides of the head. Theoretically, such information might help in locating blockages in the internal carotid system, which supplies blood to the brain.

Another special type of plethysmograph is the *oculo pneumo plethysmograph*, shown in Figure 6.46. As the name implies, this instrument measures every minute volume changes that occur in the eye with each

Figure 6.46. Oculo pneumo plethysmograph. (Courtesy of Electro-Diagnostic Instruments, Burbank, CA.)



arterial blood pulsation. A small eye cup is placed over the sclera of each eye and is connected to a transducer positioned over the patient's head by a short section of flexible tubing. A vacuum, which can be varied from zero to −300 mm Hg, is applied to hold the eye cups in place. Pulsations are recorded on two channels of a three-channel pen recorder, one for each eye. The third channel is used to record the vacuum. By periodically allowing the vacuum to build up to −300 mm Hg and deplete to zero, the instrument can also be used as a recording suction *ophthalmodynamometer*, an instrument for measuring arterial blood pressure within the eye. Ocular plethysmography and blood pressure measurements are of particular interest because the eye provides a site for noninvasive access to the cerebral circulation system. Occlusion of one of the internal carotid arteries or other interference in the blood supply to one hemisphere of the brain can be detected by nonsymmetric pressures and pulsations at the eyes.

In certain rodents the tail is a convenient region for measurement of circulatory factors. For these measurements, a *caudal plethysmograph* is used. Caudal plethysmographs can utilize any of the previously described methods of sensing volume changes or the presence of blood pulsations. The same limitations encountered in human plethysmographic procedures are also found in caudal plethysmography. In addition, a special physiological factor must be considered in measuring blood pulsations from the tail of a rodent. Many animals use their tails as radiators in the control of body temperature. At low temperatures, very little blood actually flows through the vessels of the tail, and plethysmographic measurements become very difficult. If the animal is heated to a temperature at which the tail is used for cooling, however, sufficient blood flow for good plethysmographic measurements is usually found. Sometimes the necessary temperature for good caudal measurements is so near the point of overheating that traumatic effects are encountered.

6.5. MEASUREMENT OF HEART SOUNDS

In the early days of auscultation a physician listened to heart sounds by placing his ear on the chest of the patient, directly over the heart. It was probably during the process of treating a well-endowed, but bashful young lady that someone developed the idea of transmitting heart sounds from the patient's chest to the physician's ear via a section of cardboard tubing. This was the forerunner of the stethoscope, which has become a symbol of the medical profession.

The *stethoscope* (from the Greek word, *stethos,* meaning chest, and *skopein,* meaning "to examine") is simply a device that carries sound energy from the chest of the patient to the ear of the physician via a column

of air. There are many forms of stethoscopes, but the familiar configuration has two earpieces connected to a common bell or chest piece. Since the system is strictly acoustical, there is no amplification of sound, except for any that might occur through resonance and other acoustical characteristics.

Unfortunately, only a small portion of the energy in heart sounds is in the audible frequency range. Thus, since the dawning of the age of electronics, countless attempts have been made to convince the medical profession of the advantage of amplifying heart sounds, with the idea that if the sound level could be increased, a greater portion of the sound spectrum could be heard and greater diagnostic capability might be achieved. In addition, high-fidelity equipment would be able to reproduce the entire frequency range, much of which is missed by the stethoscope. In spite of these apparent advantages, the *electronic stethoscope* has never found favor with the physician. The principal argument is that doctors are trained to recognize heart defects by the way they sound through an ordinary stethoscope, and any variations therefrom are foreign and confusing. Nevertheless, a number of electronic stethoscopes are available commercially.

Instruments for graphically recording heart sounds have been more successful. As stated in Chapter 5, a graphic record of heart sounds is called a *phonocardiogram*. The instrument for producing this recording is called a *phonocardiograph*. Although instruments specifically designed for phonocardiography are rare, components suitable for this purpose are readily available.

The basic transducer for the phonocardiogram is a microphone having the necessary frequency response, generally ranging from below 5 Hz to above 1000 Hz. An amplifier with similar response characteristics is required, which may offer a selective lowpass filter to allow the high-frequency cutoff to be adjusted for noise and other considerations. In one instance, where the associated pen recorder is inadequate to reproduce higher frequencies, an integrator is employed and the envelope of frequencies over 80 Hz is recorded along with actual signals below 80 Hz.

The readout of a phonocardiograph is either a high-frequency chart recorder or an oscilloscope. Because most pen galvanometer recorders have an upper-frequency limitation of around 100 or 200 Hz, photographic or light-galvanometer recorders are required for faithful recording of heart sounds. Although normal heart sounds fall well within the frequency range of pen recorders, the high-frequency murmurs that are often important in diagnosis require the greater response of the photographic device.

Some manufacturers of multiple-channel physiological recording systems claim the phonocardiogram as one of the measurements they offer. They have available as part of their system a microphone and amplifier suitable for the heart sounds, the amplifier often being the same one used for EMG (see Chapter 10). Some of these systems, however, have only a pen

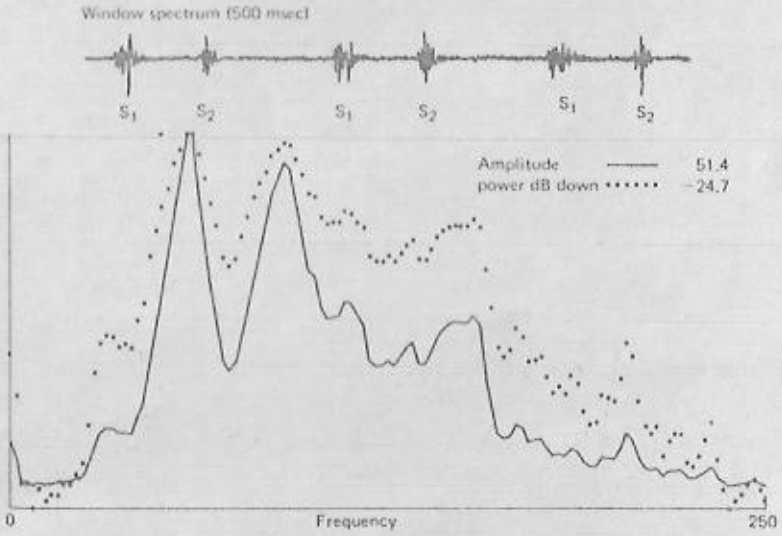
recorder output, which limits the high-frequency response of the recorded signal to about 100 or 200 Hz.

The presence of higher frequencies (murmurs) in the phonocardiogram indicates a possible heart disorder. For this reason, a spectral analysis of heart sounds can provide a useful diagnostic tool for discriminating between normal and abnormal hearts. This type of analysis, however, requires a digital computer with a high-speed analog-to-digital conversion capability and some form of Fourier-transform software. A typical spectrum of heart sounds is shown in Figure 6.47.

Microphones for phonocardiograms are designed to be placed on the chest, over the heart. However, heart sounds are sometimes measured from other vantage points. For this purpose, special microphone transducers are placed at the tips of catheters to pick up heart sounds from within the chambers of the heart or from the major blood vessels near the heart. Frequency-response requirements for these microphones are about the same as for phonocardiograph microphones. However, special requirements dictated by the size and configuration of the catheter must be considered in their construction. As might be expected, the difference in acoustical paths makes these heart-sound patterns appear somewhat different from the usual phonocardiogram patterns.

The *vibrocardiograph* and the *apex cardiograph*, which measure the vibrocardiogram and apex cardiogram, respectively, also use microphones

Figure 6.47. Frequency spectrum of heart sounds. (Courtesy of Computer Medical Science Corporation, Tomball, TX.)



as transducers. However, since these measurements involve the low-frequency vibrations of the heart against the chest wall, the measurement is normally one of displacement or force rather than sound. Thus, the microphone must be a good force transducer, with suitable low-frequency coupling from the chest wall to the microphone element. For the apex cardiogram, the microphone must be coupled to a point between the ribs. A soft rubber or plastic cone attached to the element of the microphone gives good results for this purpose.

Because the vibrocardiogram and the apex cardiogram do not contain the high-frequency components of the heart sounds, these signals can be handled by the same type of amplifiers and recorders as the electrocardiogram (see Section 6.1). Often, these signals are recorded along with a channel of ECG data to maintain time reference. In this case, one channel of a multichannel ECG recorder is devoted to the heart vibration signal.

For recording the Korotkoff sounds from a partially occluded artery (see Chapter 5 and Section 6.2), a microphone is usually placed beneath the occluding cuff or over the artery immediately downstream from the cuff. The waveform and frequency content of these sounds are not as important as the simple identification of their presence, so these sounds generally do not require the high-frequency response specified for the phonocardiogram. Circuitry for identification of these sounds is included in certain automated, indirect blood pressure measuring devices (see Section 6.2.2).

Measurement of the ballistocardiogram requires a platform mounted on a set of extremely flexible springs. When a person lies on the platform, the movement of his body in response to the beating of his heart and the ejection of blood causes similar movement of the platform. The amount of movement can be measured by any of the displacement or velocity transducers or accelerometers described in Chapter 2.

7

Patient Care and Monitoring

One area of biomedical instrumentation that is becoming increasingly familiar to the general public is that of patient monitoring. Here electronic equipment provides a continuous watch over the vital characteristics and parameters of the critically ill. In the coronary care and other intensive-care units in hospitals, thousands of lives have been saved in recent years because of the careful and accurate monitoring afforded by this equipment. Public awareness of this type of instrumentation has also been greatly increased by its frequent portrayal in television programs, both factual and fictional.

Essentially, patients are monitored because they have an unbalance in their body systems. This can be caused by a heart attack or stroke, for example, or it may be the result of a surgical operation, which can drastically disturb these systems. By continual monitoring, the patient problems can be detected as they occur and remedies taken before these problems get out of hand.

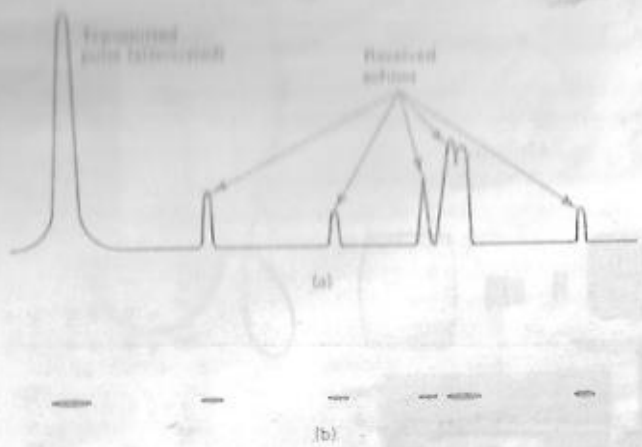
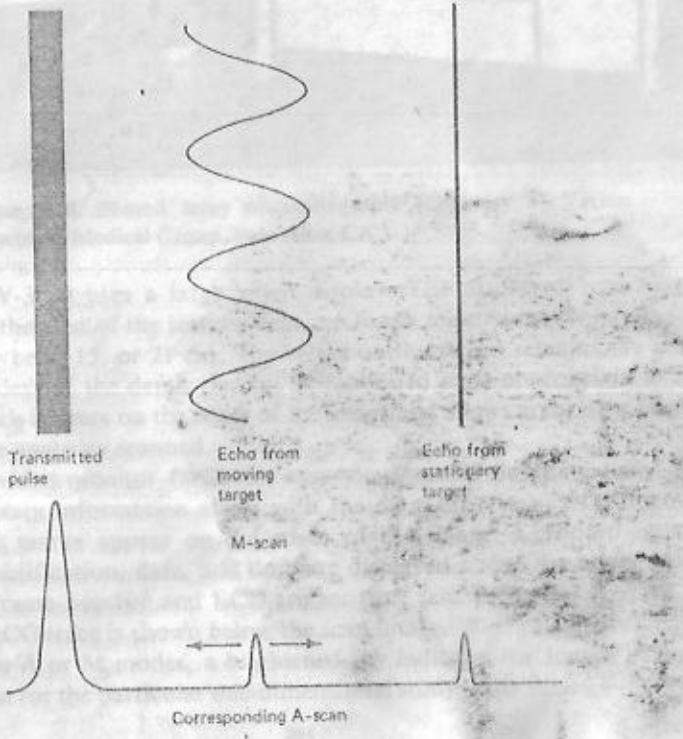


Figure 9.8. Ultrasound Display Principles. (a) Typical A-scan. Echoes cause vertical deflection of oscilloscope pattern. (b) Corresponding display in which echoes control brightness of oscilloscope beam. This principle is used in both B- and M- scan displays.

Figure 9.9 M-scan of moving and stationary target with corresponding A-scan.



trace lies immediately adjacent to the one preceding it, the dot representing each target will trace a line on the paper as shown in Figure 9.9. A stationary target will trace a straight line, whereas a moving target will trace the pattern of its movement with respect to time. A light-pen recorder in which the intensity of the light source can be controlled may be used instead of an oscilloscope to produce a chart record of the movement of echoes with respect to time. An example of an M-scan recording is the echocardiogram shown in Figure 9.12.

3. *B-scan display:* While the M-scan is used to display the movement of targets with respect to time, the B-scan presents a two-dimensional image of a stationary organ or body structure. As in the M-scan, the brightness of the oscilloscope or light-pen beam is controlled by returning echoes; however, in the B-scan the transducer is moved with respect to the body while the vertical deflection of the oscilloscope or movement of the chart paper is made to correspond to the movement of the transducer. The movement may be linear, circular, or a combination of the two, but where it is anything other than linear, the sweep must be made to compensate for the variations in order to provide a true two-dimensional display of the segment being scanned. Examples of B-scan displays are shown in Figure 9.15.

9.3. ULTRASONIC DIAGNOSIS

The applications of ultrasonic methods to medicine are many and varied. The techniques are used in cardiology, for abdominal imaging, in brain studies, in eye analysis, and in obstetrics and gynecology. The records obtained have various names, which usually include the words "echo" or "sono." For example the *echocardiogram,* analogous to the electrocardiogram, is a record of ultrasonic measurements in the heart. The *echoencephalogram* is a record obtained from the brain. A general term used, especially with eye analysis, is the *ultrasonogram.* Examples of these will be presented later in this section. Multiple names have been coined for the instruments used, including many trade titles. Typical designations are *ultrasonograph, ultrasonoscope,* and *sonofluoroscope.* The latter is used essentially for moving structures, whereas some of the devices are used for static images.

Before discussing some of the components and specific medical fields, perhaps a general overview and illustration will give the reader a perspective. Figure 9.10 is a *phased-array ultrasonograph,* which represents a class of instruments that has recently appeared. It is virtually an entire medical ultrasound laboratory in one mobile instrument.

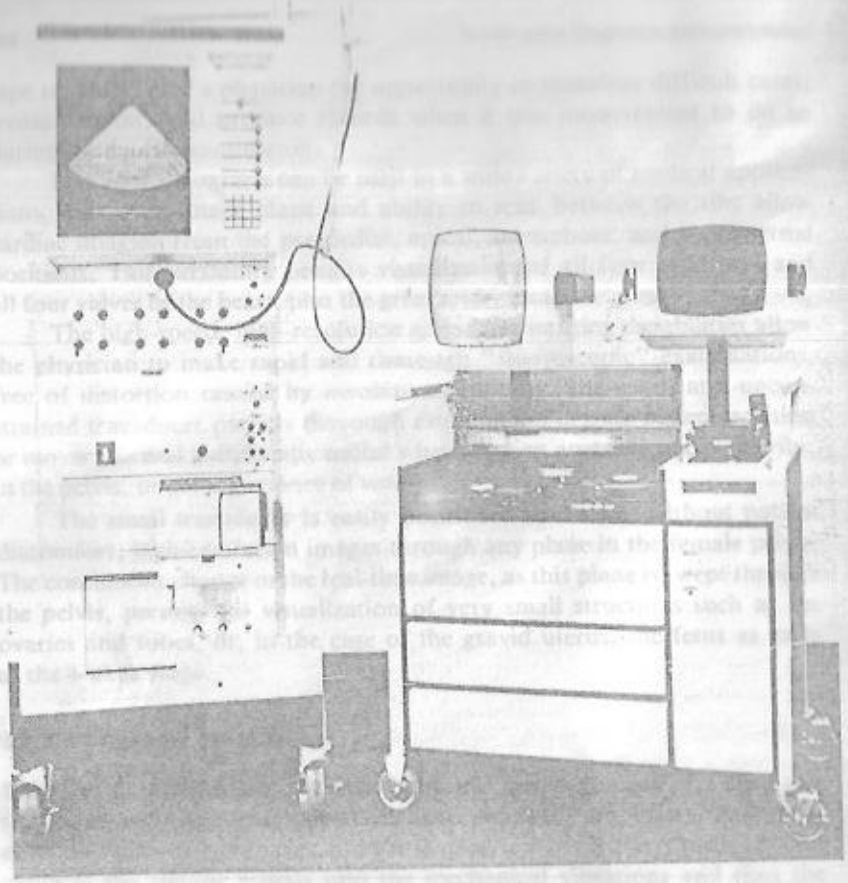


Figure 9.10. Phased array ultrasonograph. (Courtesy of Varian Associates, Medical Group, Palo Alto, CA.)

The V-3000 uses a large video display. The displayed scan image represents the area of the sector swept, up to the selected maximum depth, which may be 7, 15, or 21 cm. The image on the screen remains the same size regardless of the depth chosen. A calibrated scale appropriate to the chosen depth appears on the edges of the image and allows accurate measurement of the anatomy scanned.

The video monitor facilitates easy viewing and simultaneously presents ancillary information along with the diagnostic image. Orthogonal calibration marks appear on the edges of the image. Keyboard-entered patient identification, date, and time are displayed above the image along with the frame number and ECG trigger time relative to the R wave. A nonfade ECG trace is shown below the scan image. When the instrument is used in the A or M modes, a brightened ray indicates the spatial orientation chosen for the particular one-dimensional study being done.

Two separate sets of gain controls give maximum flexibility in optimizing the image for a specific examination. In addition to an overall gain control, a set of seven slide controls allows independent adjustment of gain for each depth interval, corresponding to one-seventh of the selected total penetration depth. Other controls are available which adjust gray scale and suppress noise.

M-mode and A-mode are available on the V-3000 in addition to the two-dimensional imaging capability. In a pure two-dimensional imaging application, the V-3000 operates at a scanning speed of 30 frames per second. For an M-mode or A-mode study, the frame rate is halved to 15 frames per second, with every other pulse used to accumulate data for the M- or A-mode. In M- or A-mode scanning, a brightened line is superimposed on the main display indicating the single ray along which the data are obtained. In the case of A-mode, these data are displayed directly along the brightened ray; for the M-mode, it appears on the optional slow oscilloscope. When a footswitch is depressed, the M-mode information is printed out on a strip-chart recorder.

For cardiac applications, an optional ECG amplifier is available. The trace appears on all displays, including the strip-chart recorder. ECG-triggered photographs may be made by positioning a cursor at the desired time position on the ECG trace.

The V-3000 includes computer-directed self-diagnosis, activated by a front-panel control. These diagnostic routines test the various circuit boards and show the results on the main display.

Phased-array sector scanning describes a technique in which the ultrasonic beam is electronically swept through an arc to produce sharp, high-resolution images in real time while the small transducer is held stationary in any desired position. Scanning is accomplished by pulsing individual crystals in the transducer at slightly different times under the control of a microcomputer. Because the emitted (and received) ultrasonic beam is swept electronically, a wide-angle image is instantly produced wherever the transducer is placed, allowing thorough and rapid imaging with no constraints on transducer positioning.

The video recorder cart is an essential element of the V-3000 system. It includes all the necessary components to do dynamic recording and allows off-line detailed study of these recordings. During an examination, video recordings can be activated by a foot switch. The cart contains a video tape recorder with slow- and stop-motion capabilities, a viewing monitor, and a camera directed at an internal live video monitor optimized for photography.

The video recorder cart is a complete video playback system, and is easily disconnected from the diagnostic module. It can be moved to remote locations for video-tape viewing and photographic record making. The off-line photographic capabilities, combined with the stop action of the video

tape recorder, give a physician the opportunity to reanalyze difficult cases, freeze motion, and produce records when it was inconvenient to do so during the actual examination.

This ultrasonograph can be used in a wide variety of medical applications. The wide image plane and ability to scan between the ribs allow cardiac imaging from the precordial, apical, subxiphoid, and suprasternal positions. This versatility permits visualization of all four chambers and all four valves of the heart, plus the great arteries and the great veins.

The high-speed, high-resolution gray-scale imaging capabilities allow the physician to make rapid and thorough "fluoroscopic" examinations, free of distortion caused by involuntary motion. The small and unconstrained transducer permits thorough examination despite patient position or movement and is especially useful when imaging anatomy under the ribs, in the pelvis, or in the presence of sutures.

The small transducer is easily positioned to obtain, without patient discomfort, high-resolution images through any plane in the female pelvis. The continuous change in the real-time image, as this plane is swept through the pelvis, permits the visualization of very small structures such as the ovaries and tubes, or, in the case of the gravid uterus, the fetus as early as the 4-week stage.

9.3.1. Ultrasonic Transducers

Whatever the application, the basic ultrasonic system consists of a generator for the electric signal, a transducer, the necessary amplifiers, and other electronic processing devices and the display unit. It is the transducer that converts the electric signals into the mechanical vibrations and thus the acoustic waves which form the basis of the technique. Transducers are produced in a variety of configurations for the various applications, and with varying frequency capabilities. The acoustic wave from the transducer enters the body through the skin surface and is then propagated in a predetermined beam pattern (wide or narrow depending on requirements) toward the structure to be examined. When the ultrasonic waves strike an acoustic interface such as the boundary of an organ, some energy is reflected. This reflected energy is picked up on the transducer and is amplified, processed, and finally displayed on an oscilloscope.

Figure 9.11 shows a Dapco echocardiology transducer. These are available in a variety of internal focal arrangements. In transducers of this type, the length of the focal zone is determined by the distance from the transducer to the cardiac structures of interest. The choice of frequency depends on the amount of tissue attenuation encountered. Transducers are available at frequencies of 1.6, 2.25, 3.5, and 5.0 MHz and with crystal diameters of 6, 13, and 20 mm for each frequency.

Figure 9.11. Echocardiology transducer. (Courtesy of Dapco Industries, Ridgefield, CT.)

The size of a transducer generally refers to the size of the active element (diameter in millimeters). The size is determined by the area to be examined. In the case of general abdominal B-scanning, there are few anatomical restrictions. However, a larger-diameter transducer provides ease of mechanical scanning and optimum focus at the required longer focal zones. Smaller transducers are best suited for investigations in anatomic areas of irregular shapes, reduced skin surface, and investigations necessitating improved resolution with minimal penetration (i.e., eye, neck, chest, and limbs). In echocardiography, the smaller active-element diameters are used to facilitate ease of placement and maneuverability within the intercostal space. The 13-mm active-element diameter transducer is most commonly used for investigations on the average adult. When investigating an extremely large and/or obese patient or when placement is not critical, a 20-mm diameter can be used. On the other hand, a 6-mm diameter is recommended for pediatric and neonatal echocardiography.

The type of tissue and the amount of tissue attenuation encountered during a diagnostic examination determines the frequency to be used. A frequency of 2.25 MHz is the most commonly used for general-purpose ultrasonography. Transducers of higher frequency, such as 3.5 or 5.0 MHz, are utilized when tissue penetration is easily accomplished and improved resolution at short distances is required. In view of the small size and shorter examination depth in children, a higher frequency is usually used to ensure good resolution. The 5.0-MHz transducer is recommended for the majority of pediatric echocardiography, reserving the 3.5 MHz for investigations of the older and larger children.

When tissue penetration becomes a concern, as with most obese patients, a low-frequency (i.e., 2.25 or 1.6 MHz) transducer should be employed, which can provide more information than can be obtained with a transducer of a higher frequency.

Echocardiology transducers are available nonfocused or in a choice of focal configurations. Generally, a focused transducer is preferred for echocardiography. The sound beams generated in a focused transducer have a reduced width to provide optimal lateral resolution and sensitivity at given depths. In selecting a focal length, two factors must be considered: the distance from the transducer to the structure to be investigated and the transducer's diameter. The focal zone of a transducer refers to the distance (in centimeters) the sound beam travels in water to a standard test object. The focal zones are commonly referred to as short (5.0 cm), medium (7.5 cm), and long (10.0 cm) in water. If echocardiography is to be performed on a small child, the distance from the transducer to the cardiac structure usually falls within 5.0 cm; therefore, a short focal length should be utilized.

Size of the transducer also affects the choice of focal lengths. For instance, a 20-mm active-element diameter can be focused best at the longer focal lengths, whereas a 13-mm or smaller diameter can accommodate all three with optimum results.

A variation of the standard echocardiology transducer is the cardiac suprasternal notch transducer. This type of transducer is designed with a small active-element diameter and is commonly utilized at a frequency of 3.5 or 5.0 MHz. When these transducers are positioned in the region of the suprasternal notch, simultaneous measurement of the aortic arch, right pulmonary artery, and left atrium can be easily obtained. The cardiac suprasternal notch transducer is extremely useful in diagnosing and monitoring a variety of heart diseases and defects, including mitral insufficiency, patent ductus arteriosus, ventricular septal defects, and hypertensive cardiovascular heart disease.

The transducers are available also as general scanners, standard medical transducers which can be utilized for A-mode or M-mode application, biopsy transducers for tissue biopsy under direct vision, plus special types for thyroid examination and ophthalmic ultrasonography. There are also microprobe needle transducers.

9.3.2. Echocardiography

One use of ultrasound in the cardiovascular system has already been discussed in Section 6.3. This is the Doppler technique in blood flow measurement. Pulsed Doppler ultrasound can be used to measure the velocity gradient across a blood vessel as well as the velocity of the heart wall or specific valves in the heart. Some additional applications will be discussed

later in the chapter. The major application in cardiovascular diagnosis, however, is the *echocardiogram,* which utilizes an M-scan technique. In the echocardiogram movements of the valves and other structures of the heart are displayed as a function of time and usually in conjunction with an electrocardiogram.

Over the past several years, echocardiology has been extremely useful in diagnosing many cardiac abnormalities, among them are calcific aortic stenosis, pulmonary valve stenosis, mitral valve stenosis, left atrial myxoma, mitral valve prolapse, and rheumatic heart disease.

In selecting a transducer for an echocardiographic investigation, the following factors must be considered for optimum results:

1. The type of investigation to be performed.
2. Physical size of the patient.
3. The anatomic area involved.
4. The type of tissue to be encountered.
5. The depth of the structure(s) to be studied.

Figure 9.12 is a typical echocardiogram. Figure 9.13 is a sketch showing placement of the transducer. For this particular echocardiogram the transducer was placed so that the beam crossed the chest wall into the right ventricle, through the septum, into the left ventricle, and ultimately through the left atrium. The aorta and mitral valve are also imaged.

With ultrasound it is possible to distinguish between different soft tissues and to measure the motion of structures of the heart. This fact has made it a valuable method of analysis in cardiology. One important factor is that there is virtually no interference by echoes from other body structures, since the heart is surrounded by the lungs, which are literally air bags. This helps greatly in the interpretation. The heart has a number of acoustic interfaces, such as the atrial and ventricular walls, the septum, and the various valves. The position and movements of each interface can be measured by the reflected ultrasound. The echoes from these walls and valves are predictable since the components of the heart move in a known manner.

A good example of this technique is shown in Figure 9.12. This type of echocardiogram is useful in interpreting the movements of the mitral valve with respect to time. The mobility of the valve is measured by the displacement of the echo per unit time during diastole. If the mobility is reduced, as in the case of mitral stenosis (narrowing of the left ventricular orifice), its severity compared with other existing conditions can determine the appropriate action to take, including the possible use of surgery.

Another use of echocardiography is in the detection of fluids. When the pericardium (the sac surrounding the heart) is inflamed (a condition known as pericarditis), there is sometimes an escape of fluid. The presence of this fluid can be detected in the echocardiogram.

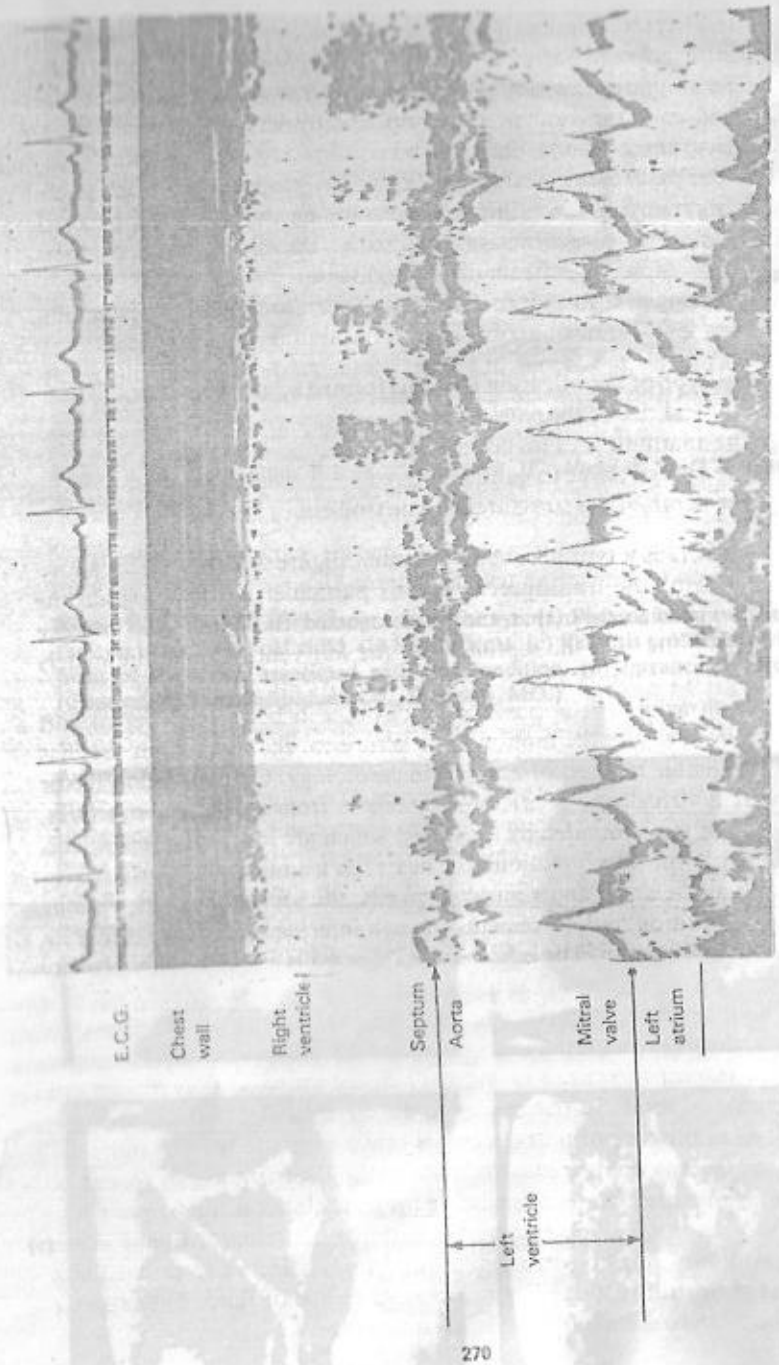


Figure 9.12. Echocardiogram. (Courtesy of Dept. of Cardiology, Cedars-Sinai Hospital, Los Angeles, CA.)

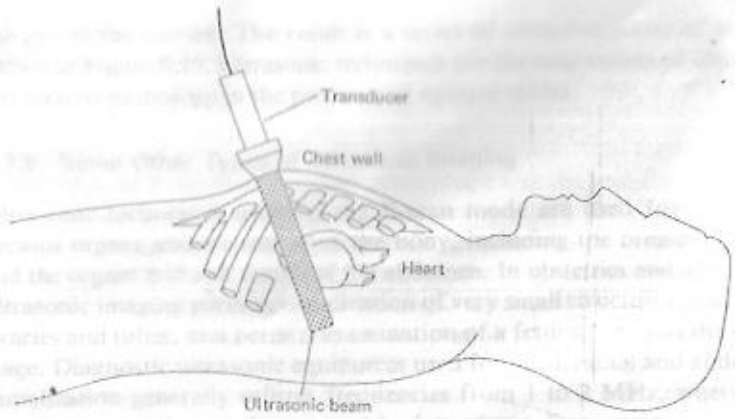


Figure 9.13. Typical transducer placement to obtain echocardiogram.

Fast scanning speeds are required to prevent blurring of the image due to heart movements. To give a thorough dynamic analysis, many simultaneous recordings can be taken in different positions so as to give cross-sectional images at various points along the length of the heart. As techniques and interpretations continue to advance, the medical profession believes that the potential of this diagnostic method will continue to increase.

9.3.3. Echoencephalography

A well-established clinical application of ultrasonic imaging using the A-scan mode of display is the echoencephalogram, which is used in determining the location of the midline of the brain. An ultrasonic transducer is held against the side of the head to measure the distance to the midline of the brain. The midline echoes from both sides of the head are simultaneously displayed on the oscilloscope, one side producing upward deflection of the beam and the other producing downward deflection.

In the normal brain these two deflections line up, indicating equal distance from the midline to each side of the head. Nonalignment of these deflections indicates the possibility of a tumor or some other disorder that might cause the midline of the brain to shift from its normal position. The instrument for this measurement produces ultrasound energy at a frequency of from 1 to 10 MHz. The pulse rate is 1000 per second.

9.3.4. Ophthalmic Scans

Another important application involves diagnostic scanning of the eye. Figure 9.14 shows an ophthalmic B-scan with camera. The transducer is shown at the bottom of the picture. The transducer is placed directly over

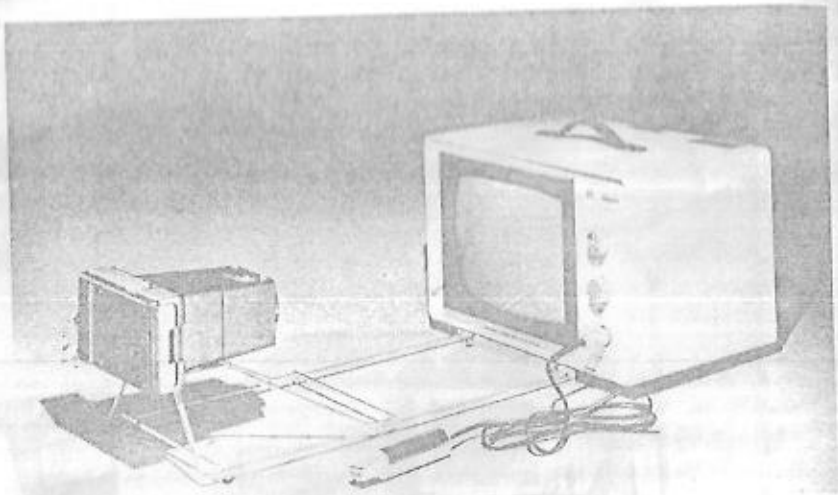
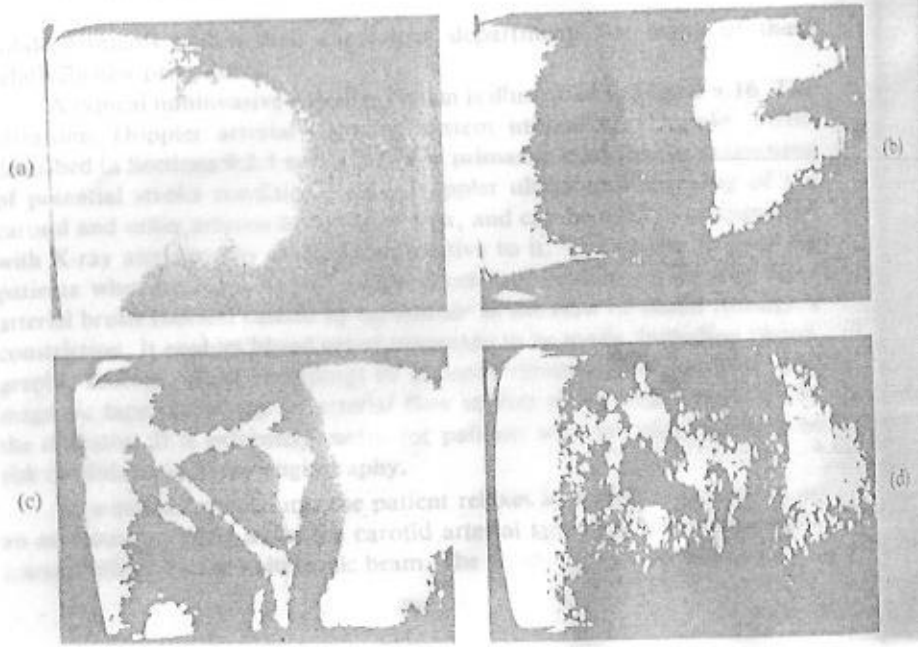


Figure 9.14. Ophthalmic B-scan with camera. (Courtesy of Storz Instrument Co., St. Louis, MO.).

Figure 9.15. Typical ultrasonograms of the eye. (a) Phthisis bulbi (wasting away of the eye) in a 13-year-old child. (b) Papilledema. (c) Retinitis proliferans with detachment of the retina associated with this condition. (d) Intraocular foreign body. (Courtesy of Storz Instrument Co., St. Louis, MO.)



the eye of the patient. The result is a series of ultrasonograms of the type shown in Figure 9.15. Ultrasonic techniques are the only means of identifying intraocular pathology in the presence of opaque media.

9.3.5. Some Other Types of Ultrasonic Imaging

Ultrasonic techniques utilizing the B-scan mode are used for visualizing various organs and structures of the body, including the breasts, kidneys, and the organs and soft tissues of the abdomen. In obstetrics and gynecology, ultrasonic imaging permits visualization of very small structures, such as the ovaries and tubes, and permits examination of a fetus as early as the 4-week stage. Diagnostic ultrasonic equipment used for intracranial and abdominal visualization generally utilizes frequencies from 1 to 2 MHz, whereas for examination of the eye, breast, and body surfaces, frequencies in the range 4 or 5 to 15 MHz are used to obtain better resolution.

9.3.6. Other Applications

Ultrasonic tomography techniques, in which information from scans taken from many different vantage points are combined mathematically, provide increased detail in the visualization of certain parts of the body. The principle involved is very similar to that of computerized axial tomography utilizing X-ray information, which is described in detail in Chapter 15. However, because of the limited penetrating range, particularly at higher frequencies, and the requirement that no gaseous regions or bone lie in the ultrasonic path, the uses in which ultrasonic tomography is practical are limited. Where these techniques can be used, however, they provide a way of obtaining detailed cross sections without radiation exposure.

9.3.7. The Noninvasive Vascular Laboratory

In Section 9.3.2 the diagnosis of the heart and the use of ultrasound in cardiology were discussed. In Section 6.3.2, measurement of blood flow by ultrasonic methods was described. Also, in the introductory paragraphs of Section 9.3, the phased-array ultrasonograph was presented as an example of a medical ultrasound laboratory. To expand on these ideas and to help complete the picture on the status of the use of ultrasound in cardiology, it is appropriate to introduce the *noninvasive vascular laboratory*. Although defined in many ways and by various manufacturers of the necessary equipment, this type of laboratory, in general, involves the scanning of the vascular portion of the cardiovascular system, and, as in inferred by the name, all methods are noninvasive. Also, the procedures can be performed with outpatients as well as those in the hospital. Some major hospitals have set up

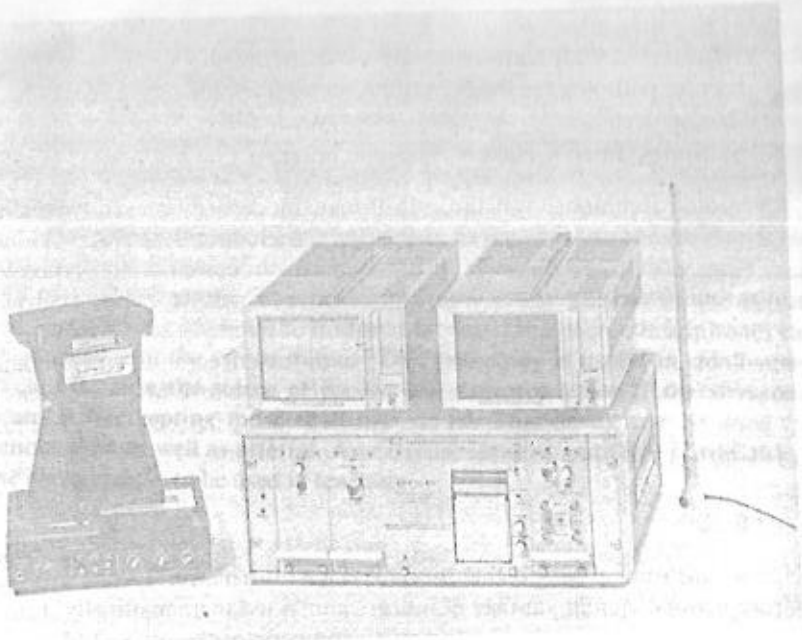


Figure 9.16. Dopscan Ultrasonic Doppler arterial scanning system. (Courtesy of Carolina Medical Electronics, King, NC.)

subdepartments within their cardiology departments for many of these relatively new procedures.

A typical noninvasive vascular system is illustrated in Figure 9.16. The ultrasonic Doppler arterial scanning system utilizes the Doppler effect described in Sections 9.2.1 and 9.2.2. It is primarily used for the diagnosing of potential stroke conditions using Doppler ultrasound scanning of the carotid and other arteries in the neck area, and can be used in conjunction with X-ray angiography or as an alternative to it. The system is used for patients who are suspected of reduced cerebral circulation or who have arterial bruits (noises) caused by turbulence in the flow of blood through a constriction. It enables blood vessel mappings to be made, including photographic records, chart recordings of pulsatile directional blood flow, and magnetic tape recordings of arterial flow sounds along with comments by the operator. It is extremely useful for patients who are considered to be risk candidates in X-ray angiography.

In a normal procedure, the patient relaxes in a reclining position on an examination table while the carotid arterial system is transcutaneously scanned with a focused ultrasonic beam. The small, smooth probe is attached

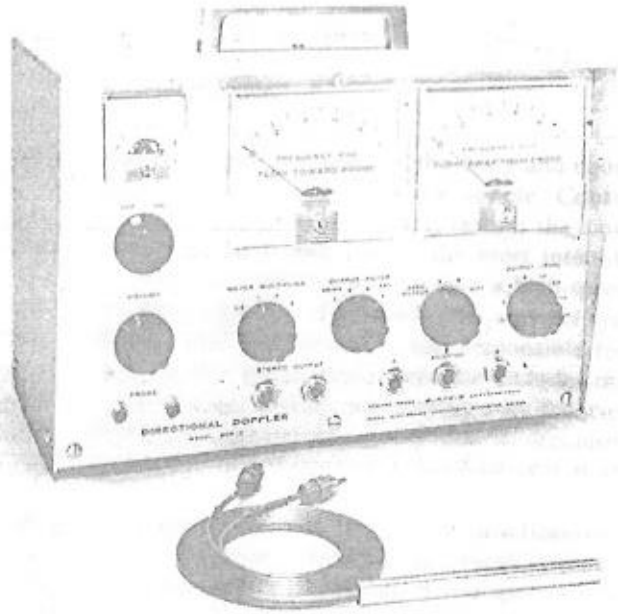
to an X-Y position-sensing arm. As the technician moves the probe in a prescribed scanning pattern over the skin, the probe's position is plotted on the screen of a storage oscilloscope whenever it senses blood flow. Repeated passes over the carotid bifurcation and adjacent arteries gradually construct on the oscilloscope a representation of the vessels scanned. Simultaneously, blood flow velocity is displayed on a monitor scope and the sounds characteristic of ultrasonically detected blood flow are available by means of a loudspeaker or headphones. All information is permanently recorded.

When the screening is complete, the vascular map is photographed, and this, along with the pulsatile flow tracings, the sound recordings, and the operator's comments provide permanent graphic and aural records for subsequent clinical evaluation.

Although the image obtained does not have as much intimate detail as one obtained by angiography, it is quite adequate in most cases. For example, it is quite easy to detect the narrowing of a major artery (arterial stenosis).

This system is also useful in investigating blood flow in the ophthalmic region and some of the pathways around the vertebrae and the base of the skull. It can be used on other major superficial vessels such as femoral, brachial, and popliteal arteries.

Figure 9.17. Directional Doppler. (Courtesy of Parks Electronic Laboratory, Beaverton, OR.)



Another useful device for the noninvasive vascular laboratory is illustrated in Figure 9.17. This instrument is a single-frequency directional Doppler blood flow detector. A dual-frequency model is also available. The single-frequency model employs either 5 or 10 MHz, whereas the dual-frequency type has both frequencies available in the same machine. The 10-MHz ultr .sound is better for the small arteries around the eye, whereas the lower frequency gives better results for the deep vessels of the thigh and the iliacs.

Examining the face of the instrument in the photograph, the directional aspects can be observed on the two meters showing blood flow away from and toward the probe. The probe is shown at the bottom of the picture and is plugged in when used. It should be noted that stereo headphones can be plugged into the stereo output. This procedure is useful in small-vessel studies because the stereo effect lets the operator know if the ultrasonic beam is intercepting more than one vessel. This device can be used for venous flow as well as arterial. An external speaker-amplifier is available if the instrument is to be used in teaching.

10

The Nervous System

The task of controlling the various functions of the body and coordinating them into an integrated living organism is not simple. Consequently, the nervous system, which is responsible for this task, is the most complex of all systems in the body. It is also one of the most interesting. Composed of the brain, numerous sensing devices, and a high-speed communication network that links all parts of the body, the nervous system not only influences all the other systems but is also responsible for the behavior of the organism. In this broad sense, *behavior* includes the ability to learn, remember, acquire a personality, and interact with its society and the environment. It is through the nervous system that the organism achieves autonomy and acquires the various traits that characterize it as an individual.

A complete study of the nervous system, with all its ramifications, would be far beyond the scope of this book. However, an overall view can be given that provides the reader with a physiological background for measurements within the nervous system, as well as some understanding of